Muscle function in older people : relationship with falling and the response to training

Perry, Mark

Muscle function in older people: relationship with falling and the response to training.

Mark Christopher Perry

2005

Submitted in partial fulfilment of the requirements for the Degree of Doctor of Philosophy of the University of London

Division of Applied Biomedical Research

School of Biomedical Sciences

King's College London

Acknowledgements

Thanks to:

Professor Di Newham, Serena Carville, Dr Christopher Smith, Dr Olga Rutherford,

Professor Roger Woledge, Dr Iain Beith, Dr Caroline Alexander, Dr Margaret Mayston,

Mark Davis, the European Union "Better Ageing" group

and my family - Leela and Christopher.

ABSTRACT

Ageing causes a reduction in lower limb muscle strength and power and increases the rate and severity of falls. Strength training may improve most of these ageing effects, though not to levels in younger people. However there are many unanswered questions concerning age and musculoskeletal function.

Risk factors for medically unexplained falls may include reduced power, strength and left-right symmetry of power in the lower limbs but conflicting reports exist. There is some evidence that lower limb muscle force control (steadiness) may deteriorate with age, but this has not been measured in a functional context, nor investigated as a possible factor in falls risk. The causes of reduced steadiness with age have also not been fully investigated. The effects of strength training on asymmetry have also not been determined, and studies investigating training effects on steadiness are conflicting.

Forty-four young people (18-40 years), 44 healthy community-dwelling non-fallers (>70 years), and 34 healthy community-dwelling fallers (>70 years) were recruited. Measurements were made of isometric and isokinetic knee and ankle extensor/flexor strength, lower limb power, rectus femoris cross-sectional area, and isometric, concentric, eccentric and functional steadiness. Fallers had poorer steadiness, strength, normalised power and strength/power symmetry than non-fallers. The non-fallers were found to have poorer functional and concentric steadiness than the young, but similar strength/power symmetry.

Needle and surface electromyography was used to investigate the causes of age-related reduced steadiness in a small sample of subjects. Age changes in motor unit synchrony, firing variability and co-activation did not relate to the steadiness decline.

Older subjects were assigned to a control group or a strength-training group. The latter strength trained their quadriceps and hamstrings twice a week for 52 weeks. Sixty-four older subjects returned for re-testing after 12 months. Strength training did not influence steadiness or strength asymmetry, but decreased asymmetry of power.

In conclusion, steadiness declines with age and reduced steadiness may relate to falling, but strength training may not be an effective way to counter this. However, power asymmetry, which relates to falling, was improved by training.

Table of Contents

LIST OF FIGURES

Page No

LIST OF TABLES

LIST OF ABBREVIATIONS

↑	increased
↓	decreased
3D	three dimensional
ACSA	Anatomical cross sectional area
ANOVA	Analysis of variance
AP	adductor pollicis
ATPase	Adenosine triphosphatase
C	Concentric
CIS	Common input strength
cm	centimetre
CoV	Co-efficient of variation
CoV_{ISI}	Coefficient of variation of interspike interval
CSA	Cross sectional area
CT	Computed tomography
DF	Dorsiflexors
deg	Degree
DHP	Dihydropyridine
E	Eccentric
EEG	Electroencephalogram
EMG	Electromyography
F	Fallers
F/CSA	Force per cross sectional area
FDI	first dorsal interroseus
Fig.	Figure
FRT	Functional Reach test
g	acceleration due to gravity (9.81 ms^{-2})
GLM	General linear measures
H	Hamstrings
H/Q	Ratio of hamstring to quadriceps activation
I	Isometric
IEMG	Integrated electromyography
Iso	Isometric
k	kappa
KE	knee extensors
kg	kilogram
L	Least steady
LSD	Least significant difference
m	metre
max	maximum
MGF	Mechano growth factor
MHC	Myosin heavy chain
min	minimum
MRI	Magnetic resonance imaging
MVC	Maximum voluntary contraction
N	Newton
n	sample number
NF	Non-fallers
O	Older subjects
P	Probability of null hypothesis being true (1=certain)
PF	Plantarflexors

PF/DF	Ratio of plantarflexor to dorsiflexion activation
PC	Personal computer
PCSA	Physiological cross sectional area
Q	quadriceps
R	Pearson correlation co-efficient
rad	Radian
RF	Rectus femoris
RM	repetition maximum
RMS	root mean squared
Ry	Ryanodine
S	strong
SO	Strength training only group
STC	Strength training and Tai Chi group
SD	Standard deviation
SDo	step down
SE	Standard error
sec	Second
St	Steadiest
ST	stand
SU	step up
TUG	The Get Up and Go test
US	Ultrasound
Vmax	Maximum shortening velocity of muscle
W	Watt
We	weak
Y	Young subjects
yrs	Years

1 Introduction

Ageing leads to a decline in many human body systems (Holliday 2000), with negative effects on function (Steffen et al. 2002), morbidity (Swift 2001, Straus 2001) and mortality (Holliday 2000, Swift 2001, Straus 2001). An improved understanding of mechanisms underlying ageing therefore has the potential to improve quality of life for older people. It is estimated that by 2050, over 20% of the world's population will be over 60 years (Palacios 2002) and so this is an important issue. This thesis will focus on the effects of age on muscle function and the association with medically unexplained falls.

Although much has been learnt about how muscle function changes with age, there are many areas that have been sparingly researched. For example, whilst it is firmly established that strength and power decline with age (for review see Macaluso and de Vito 2004) there is conflicting or limited evidence about the different effects of ageing on different contraction types (Porter et al. 1997, Delbaere and Bourgois 2003) and the influence of voluntary activation (Hurley et al. 1998, Stackhouse et al. 2001) and antagonist co-contraction (Hakkinen et al. 1998a, Izquierdo et al. 1999, Macaluso et al. 2002, Tracy and Enoka 2002) on strength. The effects of ageing on the control of lower limb muscle force, or "steadiness" (which can be defined as the fluctuation in force output), have not been fully examined (Hortobgayi et al. 2001, Tracy and Enoka 2002), and measurement of steadiness during functional tasks has not been performed. Other unresolved issues are the effects of steadiness on function (Manini et al. 2005), the mechanisms for reduced steadiness with age (Laidlaw et al. 2000, Semmler et al. 2000a, Tracy and Enoka 2002) and the effects of muscle training on lower limb steadiness (Hortobgayi et al. 2001, Tracy and Enoka 2002).

One of the most devastating effects of ageing is the increased likelihood of falling. Falls occur "when a person comes to rest inadvertently on the ground or a lower level" (Alexander 2001). Falls are a particular problem for older members of society compared to younger adults, both in terms of the increased likelihood of falling and the seriousness of the consequences (Kenny et al. 2001). One third of people over the age of 60 years will fall at least once annually (Swift 2001) and half of these people may fall more than twice (Graafmans et al. 1996). After the age of 75, fall rates are even higher (Kenny et al. 2001) and women seem to be worse affected than men (Swift 2001).

Compounded with other aspects of ageing, such as reduced bone density (Kenny et al. 2001), 50% of all falls in the elderly will lead to some kind of injury (Alexander 2001) and 40% of these injurious falls will require medical attention (Reinsch et al. 1992). Of those falls requiring medical attention, 50% will include life-threatening injuries such as hip fractures or head injury (McKee et al. 1999). Falls are consequently the sixth leading cause of death in this age group (McKee et al. 1999). For survivors, falls may have longer-term consequences such as depression and anxiety (McKee et al. 1999, Scaf-Klomp et al. 2003, Alexander 2001). Such psychological sequelae may lead to reduced activity levels, which, in turn, may predispose to further falls (McKee et al. 1999, Scaf-Klomp et al. 2003, Alexander 2001). In contrast, the incidence of serious sequelae is much lower in younger people (Kenny et al. 2001).

At present, it is not fully understood why older people fall more often than younger adults. Maintaining an upright posture when static or when moving depends on the maintenance and co-ordination of three abilities: i) adequate sensory input, ii) accurate central processing and iii) effective cortical, neurological and muscular motor output (Wolfson et al. 1985, Thelen et al. 2000). Any problem with these three abilities may lead to falls due to an inability to react adequately to disturbances of postural equilibrium.

There is no adequate categorisation system for the primary causes of elderly falls in the literature, but a useful system might be based on these three deficits. Hence appropriate categories could be: 1) sensory problems, such as visual defects (Tobis et al. 1990, Kenny et al. 2001, Lord and Dayhew 2001), vestibular disorders (Murray et al. 2005) or proprioceptive disorders (Swift 2001); 2) deficits in central processing such as those caused by polypharmacy (Kenny et al. 2001), syncope (Rubenstein and Josephson 2002) or dementia (Kenny et al. 2001); or 3) motor deficits resulting from i) pathological lesions such as Parkinson's disease or stroke (Alexander 2001); ii) age-related motor problems such as muscle weakness (Kenny et al. 2001) or reduced muscle control.

Only medically unexplained falls will be considered in this thesis, and these are likely to be at least partly due to age-related motor problems. The wide issue of 'balance', which encompasses sensory and central processing abilities, as well as motor aspects, will not be studied. Tinetti et al. (1988) have shown that in many fallers a combination of risk factors seem to play a part, but there is evidence that age-related motor problems are a common contributory, or major, cause of falls in the elderly (Kenny et al. 2001). Hence a study focussed upon this isolated cause is of relevance and importance.

Lower limb motor problems that might be relevant to falling could include deficits in the following: strength, power, symmetry of strength and power and control of muscle force (steadiness). Knowledge of how these factors relate to falls risk is incomplete. Reduced power (Skelton et al. 2002) and strength (Whipple et al. 1987, Gehlsen and Whaley 1990, Lord et al. 1999, Takazawa et al. 2003) appear to be important, but there are inconsistencies in the literature (Schwender et al. 1997, Daubney and Culham 1999, Pavol et al. 2002). There are indications that asymmetry of power may relate to falling, but this issue has only been studied by one group (Skelton et al. 2002). Finally, there have been no studies to date examining the effects of steadiness on falls risk.

Strength training has been shown to improve strength and power in older people, but its effects on steadiness are not established, and its effects on asymmetry are unknown. This thesis also aims to investigate these issues.

In the following chapters, the effects of age will be examined by comparing a group of older non-fallers to a young group. Factors associated with falling will be assessed by a comparison of the same group of older non-fallers to an age-matched group of medically unexplained fallers. The effects of training will be studied on both older groups.

2 General Methodology

This chapter describes methods common to all chapters. Details on specific procedures will appear in the appropriate chapters.

2.1 General

Baseline and 12 month follow-up muscle performance tests were performed over two separate sessions, usually separated by 7 days, due to the large number of tests. Steadiness and muscle cross-sectional area were usually measured in the first session and strength and power were measured in the second session.

2.2 Subjects

Forty-four healthy younger volunteers aged 18-40 years (mean 29.3 yrs, standard error (SE) 0.6, range 19-41 yrs) were recruited from the population of King's College London students and staff (and their friends and families), through a circular email advertising the study.

Forty-four healthy community-dwelling volunteers aged >70 years (mean 75.9 $\pm$ 0.6 yrs, range 70-86 yrs) with no history of falls were recruited from the population of south east England through advertisements in local newspapers and periodicals aimed at the elderly.

Thirty-four healthy community-dwelling 'fallers' aged >70 years (mean 76.4 $\pm$ 0.8, range 70-87 yrs) were recruited from the population of south-east England through advertisements in local newspapers and periodicals aimed at the elderly, and also through four local falls clinics. 'Fallers' were defined as people having had at least one unexplained fall over the past 12 months. This criterion has been used in other studies (Whipple et al. 1987, Gehlsen and Whaley 1990, Daubney and Culham 1999, Lord et al.

1999). A fall was defined as an incident where any part of the body (other than the feet) involuntarily hit the ground or an intervening object such as a chair.

The target for sample numbers had been 50 per group, based on a sample size calculation with power of 0.8, alpha of 0.05, estimated group values of 5 $rad.sec^{-2}$ (young) and 7 $rad.sec^{-2}$ (older) and a SD of 5 in the step up functional steadiness test. The sample size calculator used can be found online at www.health.ucalgary.ca/~rollin/stats/ssize/. The step up functional steadiness test was chosen as the primary outcome measure as it was a novel and potentially functionally relevant variable. The estimated group values were conservative extrapolations based on measurements made on the author and a similarly aged colleague. These targets were not quite attained due to time constraints, as the rate of subject acquisition, particularly for fallers, was low.

Exclusion criteria for all subjects were:

- uncontrolled hypertension, ischeamic heart disease, or any other cardiovascular disorder likely to be exacerbated by maximal muscle contractions.
- musculoskeletal pathology in the lower limbs or spine that might affect performance in the test procedures.
- any neurological disorder that might affect performance in the test procedures.
- dementia.

For the young and elderly non-faller groups, there was an additional exclusion criterion:

- A history of any falls <u>except</u> those due to external causes that might lead to falls in most people (such as collisions with vehicles or slipping on ice).

For the elderly fallers, there was also one additional exclusion criterion:

- A history of falls solely due to external causes that might lead to falls in most people (such as collisions with vehicles or slipping on ice), or solely due to sensory deficits, altered levels of consciousness (due to dizziness, drugs etc.), or neurological and orthopaedic problems.

After expressing interest in the study, subjects were sent an information sheet through the post. Elderly volunteers' General Medical Practitioners were informed of their patients' participation and the exclusion criteria, and were asked to inform the author if they felt participation would be unwise. All volunteers completed a screening health questionnaire before participation and gave written informed consent. Approval for the study was obtained from the Research Ethics Committee of King's College London.

2.3 Statistical Analysis

This section applies to all analyses except the analyses described in chapter 7, and the repeated measures analyses described in chapter 9.

Data common to young, elderly fallers and elderly non-fallers were analysed together, using a General Linear Measures (GLM) analysis. This is robust to departures from normality except asymmetrical data, and so the GLM analysis was used for all variables except the voluntary activation data, for which the Kruskal-Wallis H test was used. These analyses were done with SPSS base 11.5 software (SPSS Inc., Chicago, USA.).

The GLM univariate test allows corrections to be made for any potentially confounding variables. For all group comparisons, sex was used as a co-factor in the GLM analysis, as the male to female ratio differed between all three groups. The interaction of sex and group (sex x group) should theoretically have been more appropriate as a co-factor, as this would allow for sex corrections specific to each group. However, the number of male fallers was very small (from 0-3) and thus the estimation of the sex correction factor for the fallers would be unlikely to be valid. Sex was removed from the analysis model if not found to interact with the dependent variable.

Only variables that showed significant differences between the three groups were subjected to post-hoc tests. If the interaction between sex and the dependent variable was not significant, the Dunnett post-hoc tests were used to ascertain differences for young *vs* elderly non-fallers and elderly fallers *vs* elderly non-fallers. The Dunnett tests enabled these to be appropriately corrected, without additional correction for the differences between young *vs* elderly fallers. Dunnett tests were not possible when comparing means corrected for sex. Hence if sex remained in the model, Bonferroni post hoc tests based on the dependent variable means (corrected for sex) were used to ascertain differences between group permutations, although only results for young *vs* elderly non-fallers and elderly fallers v elderly non-fallers were recorded.

To evaluate the overall differences between groups for collections of closely related variables, the average of each subject's measurements for those variables was calculated, and a GLM univariate ANOVA analysis of the averaged data was performed. Only those subjects with full data sets for the relevant variables were included.

Alternatively, if a tendency was observed for the mean of at least 6 closely related variables to be numerically greater in one group (young *vs* old or fallers *vs* non-fallers), the non-parametric Binomial test was used to assess the probability that this trend was due to chance.

Alpha was set to 0.05 for all group comparisons, group post-hoc tests, correlations of variables and to establish the significance of sex as a cofactor. In the results sections, the P values given will be the relevant post-hoc values.

Comparisons of ordinal or ratio data were generally made with Pearson Product correlation analyses. For such analyses, the data were usually split into 6 groups: male young, female young, male fallers, female fallers, male non-fallers, female non-fallers. This was to permit analyses that were not skewed by sex or group effects.

Unless stated otherwise, data are presented as mean $\pm$ standard error of the mean, and significant differences will generally be referred to simply as "differences". Similarly, "significantly" will generally be omitted when referring to values being significantly greater or lower between groups. This is because numerical differences that are not significant will not be viewed as true differences (unless they are part of a group trend that is itself significant).

3 Lower limb strength and power in healthy young and older people.

3.1 Introduction

3.1.1 Age effects on lower limb strength

Human skeletal muscle becomes weaker with age in both upper and lower limbs, and may decline more rapidly in the latter (Frontera et al. 1991, Lynch et al. 1999, Frontera et al. 2000a, Delbaere and Bourgois 2003). The decline in strength generally becomes significant at around the age of 50 years, and continues to decrease by 12-15% per decade (Larsson et al. 1979, Young et al. 1985, Frontera et al. 1991, Skelton et al. 1994). This decline appears to be universal (Skelton et al. 1994, Faulkner and Brooks 1995) occurring even in veteran athletes (Macaluso and De Vito 2004).

In the lower limbs, both cross-sectional and longitudinal studies have shown an age-related effect in the knee flexors and extensors (Larrsson et al. 1979, Anianson et al. 1983, Young et al. 1984, Young et al. 1985, Anianson et al. 1986, Frontera et al. 1991, Overend et al. 1992, Hortobagyi et al. 1995, Lindle et al. 1997, Izquierdo et al. 1999, Roos et al. 1999, Akima et al. 2000, Frontera et al. 2000a, Schiller et al. 2000, Goodpaster et al. 2001, Hughes et al. 2001, Sinaki et al. 2001, Lanza et al. 2003) and the ankle dorsiflexors and plantarflexors (Danneskiold-Samsoe et al. 1984, Vandervoort and McComas 1986, Thelen et al. 1996, Winegard et al. 1996, Porter et al. 1997, Kent Braun and Ng 1999, Hunter et al. 2000, Lanza et al. 2003).

3.1.2 Effects of contraction type, velocity of movement and sex on strength changes with age

Some researchers have reported that isometric, concentric and eccentric strength (Lindle et al 1997, Christou and Carlton et al. 2002, Delbaere and Bourgois 2003) and isometric and concentric strength (Harridge et al. 1995) decline at similar rates. In contrast, other studies (Vandervoort et al. 1990, Poulin et al. 1992, Hortobagyi et al. 1995, Porter et al. 1995, Porter et al. 1997) have reported lower age-related declines in eccentric strength.

Age effects on eccentric (Poulin et al. 1992) and concentric (Aniansson et al. 1992, Delbaere and Bourgois 2003) strength may be less at higher velocities, but Anianson et al. (1983) found that age effects on concentric strength may be greater with higher velocities. Other studies have noted no effect of contraction speed on the decline of eccentric (Lindle et al. 1997) or concentric (Anianson et al. 1983, Aniansson et al. 1986, Cunningham et al. 1987, Laforest et al. 1990, Lindle et al. 1997, Frontera et al. 2000a, Lanza et al. 2003) contractions.

There appears to be no effect of sex on age-related declines in isometric, concentric or eccentric strength (Porter et al. 1995, Lindle et al. 1997, Bellew et al. 1998, Delbaere and Bourgois 2003). Hortobagyi et al. (1995) suggested a slower decline of eccentric strength in women but this was not subject to a statistical analysis. Gallagher et al. (1997) noted a significantly lower absolute loss of total appendicular muscle mass in women but the significance of relative losses were not discussed and the decline in absolute leg muscle mass did not differ.

3.1.3 Causes of reduced strength with age

3.1.3.i Sarcopenia

The loss of muscle strength with age is partly due loss of muscle mass, termed sarcopenia. There is clear evidence of sarcopenia from computed tomography (CT) (Klitgaard et al. 1990a, Overend et al. 1992) and MRI (Jubrias et al. 1997, Klein et al. 2001) imaging. Sarcopenia occurs due to a reduction in both the number of fibres and fibre size (Lexell et al. 1988).

3.1.3.i.i Reduction in muscle fibre numbers

Reduced fibre numbers are detectable at 25 years of age, although they are not significant until the age of 50 years (Lexell et al. 1988) and appear to be inevitable (Faulkner and Brooks 1995).

Muscle fibres may be lost because individual fibres become denervated at the neuromuscular junction (Brooks and Faulkner 1990, Faulkner et al. 1995). It has been suggested that poorer muscle recovery from contraction-induced injury in older people (Brooks and Faulkner 1990) may increase susceptibility to denervation (Faulkner et al. 1995). Muscle recovery may depend on muscle stem cell activation (Goldspink and Harridge 2004, Owino et al. 2001) and there is evidence that Mechano growth factor (MGF), which is responsible for stimulating muscle stem cell activity in response to muscle damage, may be produced in smaller amounts with age in animals (Owino et al. 2001). In addition, whole motor units may be denervated as a result of the death of the anterior horn cell (Tomlinson and Irving 1977). Grounds (2002) suggested that this could result from neurotoxity, similar to that in amyotrophic lateral sclerosis.

The relative weighting of local fibre denervation and motoneurone cell death as contributors to muscle fibre loss has not been discussed in the literature. Both appear to induce a partially adaptive mechanism, whereby collateral sprouting from intact motor units permits reinnervation of some of the denervated fibres, which leads to motor units becoming larger by motor unit remodelling (Faulkner and Brooks 1995).

3.1.3.i.ii Reduction in muscle fibre size

Sarcopenia may also result from a reduction in muscle fibre size (Lexell et al. 1988) although there has been conflicting evidence (Frontera et al. 2000a). It may occur due to relative disuse or malnutrition (Grounds 2002) or through a direct effect of age on fibre size (Hughes and Schiaffino 1999). The latter may be mediated by the reduction in hypertrophic signalling factors such as IGF-1 (Hughes and Schiaffino 1999).

3.1.3.ii Reduction in force per cross-sectional area

Sarcopenia does not appear to explain all the strength loss that occurs with age. The remaining muscle may also have a decreased force per cross-sectional area (F/CSA), (or specific force, which will be used synonymously) but this is controversial. Reduced isometric F/CSA with age has been demonstrated in the quadriceps in some studies (Young et al. 1985, Klitgaard et al. 1990a, Petrella et al. 2005) but not in others (Young et al. 1984, Overend et al. 1992, Hakkinen and Hakkinen 1991). Similarly some studies have shown reduced concentric quadriceps F/CSA in older people (Overend et al. 1992, Jubrias et al. 1997) but others have not (Frontera et al. 2000b). In the hamstrings older people have been shown to have lower F/CSA concentrically, but not isometrically (Overend et al. 1992). No age effects have been noted in dorsiflexor F/CSA isometrically (Kent-Braun and Ng 1999) but plantarflexor specific force, calculated as isometric force/muscle volume, has been shown to be higher in young people (Morse et al. 2004).

With the exception of Morse et al. (2004), all of these studies measured the anatomical CSA (ACSA) rather than the true physiological CSA (PCSA). In pennate muscles such as the gastrocnemius or vastus lateralis, ACSA underestimates PCSA (Narici et al. 1999) and the degree of underestimation grows as pennation angle increases. As pennation angle is higher in hypertrophied muscle (Kawakami et al. 1993) and thus younger people (Narici et al. 2003), ACSA may be more of an underestimation of PCSA for young than older people. This was demonstrated by Klein et al. (2001) in elbow muscles. This implies that measures of F/ACSA in the young may be greater *over*estimations of true F/PCSA than those in older people, and this may explain part of the age difference observed in F/ACSA. In contrast, Narici et al. (2003) found that the ACSA to PCSA ratios in young and old did not differ. However, the co-efficient of determination between ACSA and PCSA was <0.6, suggesting that ACSA is prone to influence from other, potentially confounding, variables, and so should be used with caution (Narici et al. 2003). This may explain the variability in results across studies using ACSA.

Any decrease in F/CSA with age may be due to a combination of extrinsic and intrinsic factors. Extrinsic factors are extra-muscular, whereas intrinsic factors involve the contractile machinery of the muscle fibres.

3.1.3.ii.i *Extrinsic factors*

Voluntary activation changes

In the lower limb there have been conflicting findings, with some studies showing no age effects on voluntary activation in the quadriceps (Hurley et al. 1998, Roos et al. 1999), dorsiflexors (Vandervoort and McComas 1986, Connelly et al. 1999, Kent Braun and Ng 1999) or plantarflexors (Vandervoort and McComas 1986). Others have shown a decline in activation with age in the quadriceps (Stackhouse et al. 2001, Stevens et al. 2001) and plantarflexors (Morse et al. 2004). Jakobi and Rice (2002) hypothesised that the discrepancies between studies might be due to the older subjects requiring more practice to obtain full activation because of reduced habitual levels of activation, and that provision for this varied between studies.

Stevens et al. (2003) and Morse et al. (2004) also noted that most studies showing an age effect used trains of supramaximal stimuli at 50 and 100Hz (Stackhouse et al. 2001) and 50Hz (Stevens et al. 2001), while most of those not showing an effect used a single or double twitch interpolation technique (Vandervoort and McComas 1986, Hurley et al. 1998, Connelly et al. 1999, Kent Braun and Ng 1999, Roos et al. 1999). Exceptions are Kent Braun and Ng (1999) who did not show an age effect with trains of stimuli and Morse et al. (2004) who showed an effect with a double twitch technique.

Another possibility is that group sizes may be responsible for differing results. With the exception of Vandervoort and McComas (1986), older group sizes in the non age-effect studies were smaller than those in the studies showing an effect (a mean of 11 subjects compared to 19). There is therefore a case for further studies using larger group sizes assessing quadriceps activation levels with the single twitch interpolation technique.

Antagonist co-activation changes

Increased antagonist co-activation may lead to a reduced resultant force acting in the agonist direction. This has been observed in older people during isometric (,Izquierdo et al. 1999, Macaluso et al. 2002, Tracy and Enoka 2002), concentric (Izquierdo et al. 1999, Tracy and Enoka 2002, Hakkinen et al. 1998a) and eccentric (Tracy and Enoka 2002) knee extension. No age effects have been noted during isometric knee flexion (Macaluso et al. 2002) or isometric plantarflexion (Morse et al. 2004).

3.1.3.ii.ii Intrinsic factors

There have been reports that older single human (Larsson et al. 1997, Frontera et al. 2000b, D'Antona et al. 2003) and mouse (Gonzalez et al. 2000) fibres have reduced F/CSA, and although this has not been noted in a recent human study (Trappe et al. 2003), this suggests that intrinsic muscular factors may also be involved. Age-related reductions in cross-bridge force, by a shift in the equilibrium between cross-bridge high and low force states may reduce intrinsic strength (Phillips et al. 1991). Reduced intracellular pH or raised intracellular inorganic phosphate levels may cause this shift towards lower force states (Phillips et al. 1991) but Phillips et al. (1993c) did not find age-related differences in these factors despite age effects on intrinsic force. Reductions in cross-bridge force could also be caused by an increase in myofilament spacing (Frontera et al. 2000b). Reductions in the number rather than force of cross bridges are less likely to be a factor in reduced intrinsic strength, as intrinsic force during stretch is relatively spared (Phillips et al. 1991).

Such age changes may be hormonally mediated. Low oestrogen levels may cause cross bridges to shift from high to low force levels with an analogous action to a low pH or a high phosphate concentration (Phillips et al. 1993a, 1993b). In addition, both men and women show marked decreases in specific force when sex steroid levels are reduced (Phillips et al. 1993a), and hormone replacement therapy (HRT) reduces losses of F/CSA in women (Skelton et al. 1999, Phillips et al. 1993a).

Changes in excitation contraction coupling (for a review see Delbono 2002) may also influence the loss of intrinsic strength. It has been shown in animals that reduced Ca^{2+} release from the sarcoplasmic reticulum (SR), due to reductions in coupled dihydropyridine (DHP) and ryanodine (Ry) receptors (Renganathan et al. 1997), may reduce force through reduced numbers of active cross bridges, although Klitgaard et al. (1989) found these receptors are not reduced in humans. Reduced efficiency of calsequestrin as a ryanodine modulator has been suggested as an alternative cause of reduced Ca^{2+} release (Margreth et al. 1999). Reductions in the efficiency of the Na^{+} - K^{+} pump (De Luca et al. 1990) may lead to reductions in sarcolemmal excitability, in turn affecting excitation-contraction coupling (De Luca et al. 1990).

D'Antona et al. (2003) reported that fibre myosin concentration was directly proportional to the intrinsic force of single human fibres.

Although there are conflicting findings (Jubrias et al. 1997, Klittgard et al. 1990a) faster fibres may be intrinsically stronger (Grindrod et al. 1987, Hortobgayi et al. 1995, Larsson et al. 1997, Frontera et al 2000b) and so a decrease in the relative area of faster fibres in a muscle could reduce F/CSA. The total area of a fibre type in a muscle will depend on the numbers of each fibre and their mean area (Lexell and Downham 1992). Counts of human fibre types have shown a greater age reduction in faster than slower fibre numbers in some studies (Larsson et al. 1979, Aniansson et al. 1992) although others have observed no difference (Hortobagyi et al. 1995, Lexell et al. 1988, Aniannson et al. 1986, Klitgaard et al. 1989). There is more consistent evidence of a greater decrease in mean area with age in faster than slower fibres (Aniansson et al. 1986, Lexell et al. 1988, Klitgaard et al. 1990a, Hortobagyi et al. 1995) although Frontera et al. (2000a) did not note any difference. Overall, it appears that the overall proportion of muscle area made up of faster fibres may be decreased with age (Klitgaard et al. 1989, Lexell and Downham 1992, Trappe et al. 2003).

It should be noted that these findings may be limited by the ATPase staining technique, which can not differentiate "hybrid" fibres with more than one myosin heavy chain (MHC) isoform (Deschennes 2004). Older fibres may contain more hybrid fibres (Klitgaard et al. 1990b) and although they may stain as a non-hybrid fibre they may have different velocity characteristics (Deschennes 2004). However, findings of decreased faster MHC isoforms (Klitgaard et al. 1989, Jubrias et al. 1997) in older muscle support the overall conclusion of greater proportions of slow muscle in older muscle.

3.1.4 Power output changes with age

Power may be a more important factor in determining function than strength (Suzuki et al. 2001, Bean et al. 2002). Skelton et al. (1994) demonstrated clear decreases in maximal leg extensor power between the ages of 65 and 89 years in both sexes. Similar results have also been demonstrated in other studies (Bosco and Komi 1980, Izquierdo et al. 1999, Macaluso and de Vito 2003, Trappe et al. 2003, Runge et al. 2004, Petrella et al. 2005) which also measured power during a single movement. Other studies have found age reductions in peak lower limb power during maximal cycling tests (Bonnefoy et al. 1998, Marsh et al. 1999, Martin et al. 2000).

As power is the product of force and velocity, any losses in power may occur through losses in force or velocity. Contraction velocity decreases with age for two main reasons. Firstly, as previously discussed, there may be an increase in the proportion of slower muscle fibres. Secondly, there is evidence of intrinsic reductions in contraction speed within each type of fibre (Larsson et al. 1997, D'Antona et al. 2003) though other work has contradicted this (Trappe et al. 2003). Narici et al. (2003) have shown that fascicle length in the gastrocnemius may reduce with age, which implies a reduction in the number of sarcomeres in series and thus contraction velocity (Jones and Round 1990). At the level of the whole muscle, changes in tendon compliance may induce reductions in contraction speed (Reeves et al. 2003a). Maganaris (2001) noted that at isometric loads of >60% of maximum plantarflexion, older subjects had more compliant achilles tendons. This might reduce the maximum shortening speed of the muscle by increasing the time delay during stretching of the more elastic tendinous tissues (Reeves et al. 2003a). Narici et al. (2003) also suggested that greater co-activation with age may reduce contraction speed.

In animal studies, slower relaxation has been observed with ageing in both fibre types (Edstrom and Larsson 1987) and may result from reduced Ca^{2+} re-uptake to the SR due to poorer active transport of Ca^{2+} secondary to reduced Ca^{2+} ATPase (Larsson and Salviati 1989), a smaller SR (Larsson and Salviati 1989), or less efficient Ca^{2+} pumps

(Margreth et al. 1999). Although unimportant for single movements, relaxation speed might affect velocity during repeated movements such as those required to recover from a fall.

3.1.5 Asymmetry of strength and power in older people

Skelton et al. (2002) found that elderly fallers had greater asymmetry of power than elderly non-fallers. However, no studies have directly compared asymmetry of strength, power or muscle CSA between young and older people.

3.1.6 Implications for this study

This summary highlights that the age effects on asymmetry of strength and power are presently unknown. The effects of contraction type and speed on age related strength loss are controversial and require further study. The importance of specific force as a factor in age-related strength decline, and the influence of voluntary activation and antagonist co-activation on specific force are not fully established. Healthy young and older subjects were studied with the following hypotheses:

Healthy older subjects will have:

1. Reduced strength in the quadriceps and hamstrings.
2. Reduced symmetry of both strength and power
3. Reduced intrinsic strength
4. Reduced voluntary activation
5. Greater co-activation of antagonists
6. Relative sparing of eccentric strength
7. Relative sparing of faster eccentric and slower concentric contractions, based on the expected slower cross bridge cycling in older subjects (see discussion).

3.2 Methods

3.2.1 Subjects

The younger and older non-faller groups (see chapter 2) only were included in this part of the study. In this chapter the older non-fallers will be referred to as "older subjects".

3.2.2 Anthropometric measurements

With the subjects not wearing shoes, height and body mass were measured.

3.2.3 Muscle tests

Isometric and isokinetic strength

Isometric and isokinetic strength data for the knee extensors and flexors were collected with an isokinetic dynamometer (Kin Com version 5.30 HS3, Chattanooga Group Inc., TN, USA). Isokinetic dynamometers have been shown to be reliable, with a CoV in repeated tests of around 10% (for isokinetic knee extension/flexion at 90 and 120°.sec^{-1} in young and elderly subjects) reported for the Cybex dynamometer (Ly and Handelsman 2002). Similar CoV values for quadriceps and hamstrings in young people have been reported on the machine used in this study by colleagues at KCL (unpublished results, Appendix 1).

The order of testing for flexion/extension and side was randomised. The order of contraction types was isometric, concentric and eccentric for reasons of convenience.

For *isometric quadriceps and hamstring measurements* the subjects were seated in the dynamometer and the backrest was adjusted so that their knees were at the forward edge of the seat. They were strapped tightly into the chair around the waist and shoulders.

The axis of knee flexion was aligned with the dynamometer lever arm axis by eye and the strain gauge cuff was placed around the shin, with the lower border at the level of the upper border of the lateral malleolus (Fig. 3.1). The weight of the lower leg was measured using a built-in facility on the dynamometer for gravity correction. After several practice attempts, peak isometric forces were measured at 20-90° of knee flexion in 10° increments for both the quadriceps and hamstrings. Each contraction lasted for 3 seconds, with a minimum rest period of 5 seconds between each. Two repetitions at each angle were allowed for each muscle group.

Fig. 3.1 The Kin Com isokinetic dynamometer.

Isokinetic measurements were made in the same positions. Measurements were made at two angular velocities: 50 and 150 ° s^{-1} in random order. The testing range was 25 - 80°

knee flexion. A rest period of at least 5 seconds was allowed between each trial. A minimum of 3 trials was performed until values decreased. The first trials were regarded as practice attempts.

Forces were measured by a load cell attached to the cuff aligned to subjects' ankles, which had been calibrated using known weights, and gravity corrections were performed. The highest values of the isometric and isokinetic measurements at each condition were recorded in Newtons. For each condition of muscle and contraction angle/speed/type, the measurements from the left and right leg were redefined into weaker and stronger leg measurements. Any unilateral measurements were not included in the analysis.

In each leg the following ratios of peak force were calculated for each muscle group for each subject to permit evaluation of the differential age effects on each type of contraction:

- peak isometric force to concentric (50°.sec^{-1})
- peak isometric force to eccentric (50°.sec^{-1})
- concentric (50°.sec^{-1}) to eccentric (50°.sec^{-1})

The following ratios of peak force were also calculated for each muscle group for each subject to permit evaluation of the differential age effects on speed of contraction:

- slower (50°.sec^{-1}) concentric to faster (150°.sec^{-1}) concentric
- slower(50°.sec^{-1}) eccentric to faster (150°.sec^{-1}) eccentric

Leg extension power output

The Nottingham Power Rig (University of Nottingham Mechanical Engineering Unit, UK) (Fig. 3.2) was used to determine power output during single leg extension. This device incorporates a footplate pedal coupled to a flywheel. When subjects push on the pedal, the peak angular velocity (Vmax) of the flywheel is measured by a light sensor, and maximal power is calculated as the product of Vmax and the known angular inertial load. This device has been shown to have good test-retest reliability with a CoV of 9.4% for a sample of young and older people (Bassey and Short 1990).

The order of testing sides was randomised. Subjects sat in the semi-recumbent chair, adjusted so that the knee angle was 10° from full extension when the footplate pedal was at its end position. With the footplate retracted, the subjects extended their hip and knee as powerfully as possible, forcing the footplate forwards, until it reached the end position. A minimum of 6 trials was obtained on each leg and measurement was curtailed when two successive measurements were below the highest. The highest measurement on each leg was recorded. The left and right measurements were redefined as higher and lower measurements.

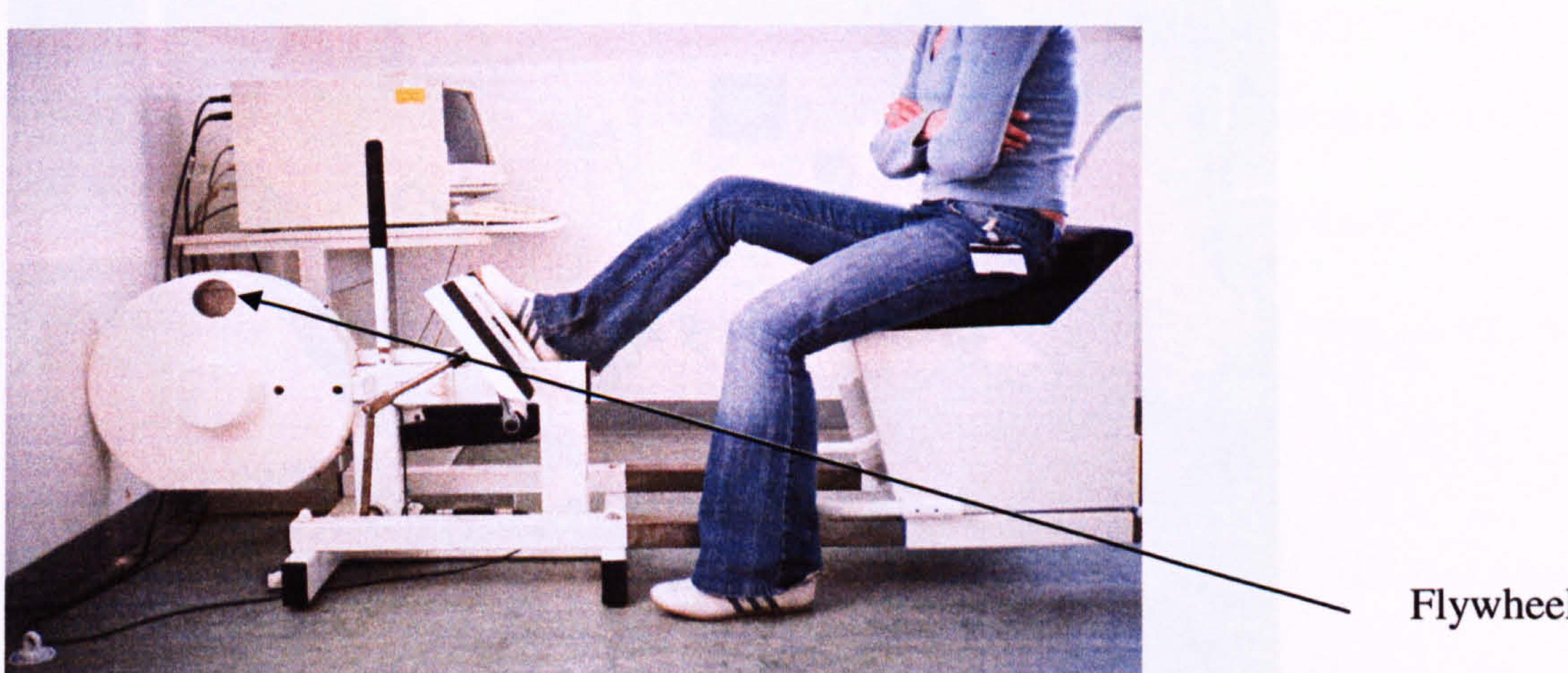


Fig. 3.2 The Nottingham Power Rig.

Rectus femoris cross-sectional area

Aloka SSd-900 ultrasound diagnostic equipment (Aloka Holding Europe AG, Switzerland) was used with an 8-cm linear transducer (frequency of 7.5 MHz) (Fig. 3.3). The cross-sectional area was measured using calibrated in-built software. Previous studies have demonstrated good reliability, with CoVs of between 4 and 4.3% in quadriceps measurement (Stokes et al. 1997).

The order of sides was randomised. The midpoint between the greater trochanter and lateral malleolus was determined using a tape measure with the subjects in supine. The transducer was placed transversely over the rectus femoris muscle belly at this point. When the muscle image was clearly visible on the screen, the image was frozen and the CSA was measured by tracing the muscle boundary on the screen using an in-built mouse. Measurements were then taken on the other leg.

To calculate an index of strength/CSA, quadriceps isometric contraction force at 80° flexion and all the quadriceps isokinetic contraction forces were divided by the CSA.

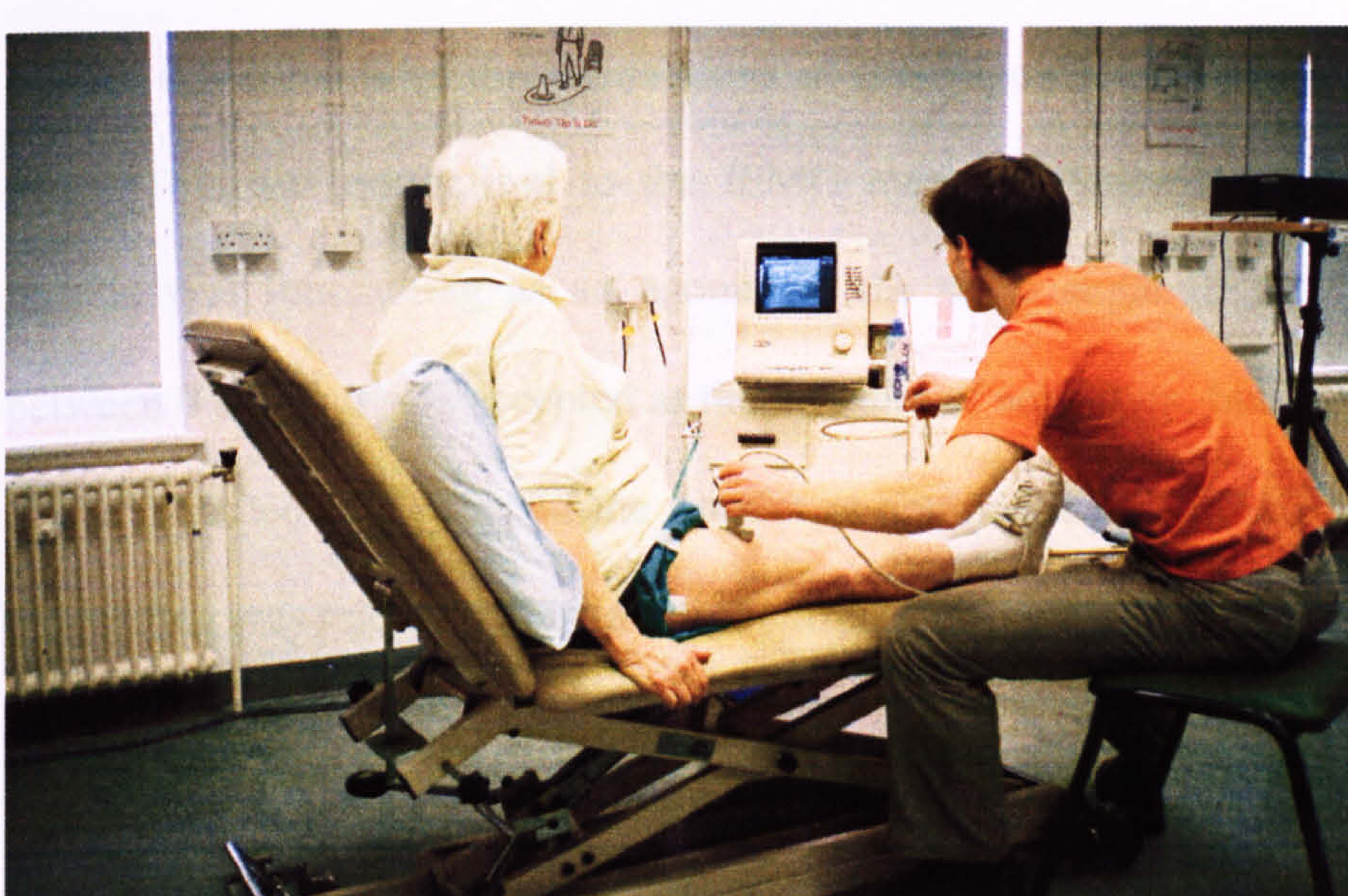


Fig. 3.3 Ultrasound measurement of RF CSA.

Asymmetry of strength, power and rectus femoris CSA.

Asymmetry was defined as the difference between sides, divided by the larger value (Skelton et al. 2002) and multiplied by 100 to yield a percentage.

Co-activation of antagonists

Surface EMG electrodes (MA-110, Motion Lab Systems, LA, USA) were used to collect integrated EMG (IEMG) data over the antagonists and agonists of both muscle groups throughout the isometric and isokinetic strength testing procedures. The electrodes contained a cathode, earth and an anode arranged in a "traffic-light" conformation. Raw EMG signals collected at 10,000Hz were fed to a filter (Neurolog NL125, Digitimer Ltd, UK) and RMS integrator (Neurolog NL705, Digitimer Ltd, UK). Low and high pass filtering occurred at 15 and 10,000Hz. Raw and integrated signals were then passed to an analogue-digital (A-D) converter (1401, CED, UK) and then to a PC, where signals were analysed using Signal 2.13 software (CED,UK).

During isometric contractions, the IEMG of the antagonists was measured during the middle 0.5 seconds of the contractions of their agonists at the agonist angle of peak force. This was normalised to the IEMG measured when the formerly antagonistic muscle was functioning as an agonist at the same angle. If this yielded a number between 0 and 1, it was accepted as the measure of activation, but if the number was >1, it was assumed that either of the agonist or antagonist measurements were faulty, and the reading was rejected. This rarely occurred.

Co-activation is the ratio of the normalised antagonist activation to the normalised agonist activation. Since the agonist activation was maximal, its normalised activation would be 1. Hence the normalised antagonist activation value was regarded as being equivalent to a co-activation value.

For isokinetic measurements, a similar system of analysis was used except that the IEMG of the antagonist at the agonist peak force angle was normalised to the peak IEMG of the formerly antagonistic muscle when acting as an agonist regardless of the angle at which it occurred.

Voluntary activation levels of the quadriceps

The subjects were strapped in an adjustable purpose-built chair with their knees at the forward edge of the seat. A strain guage was attached to a non-extensible metal cable that was attached to the subjects' ankle via a cushioned cuff. The strain guage was fitted to the back of the chair, in the centre of the transverse plane and adjusted to a height corresponding to the ankle height of the seated subjects (Fig. 3.4).

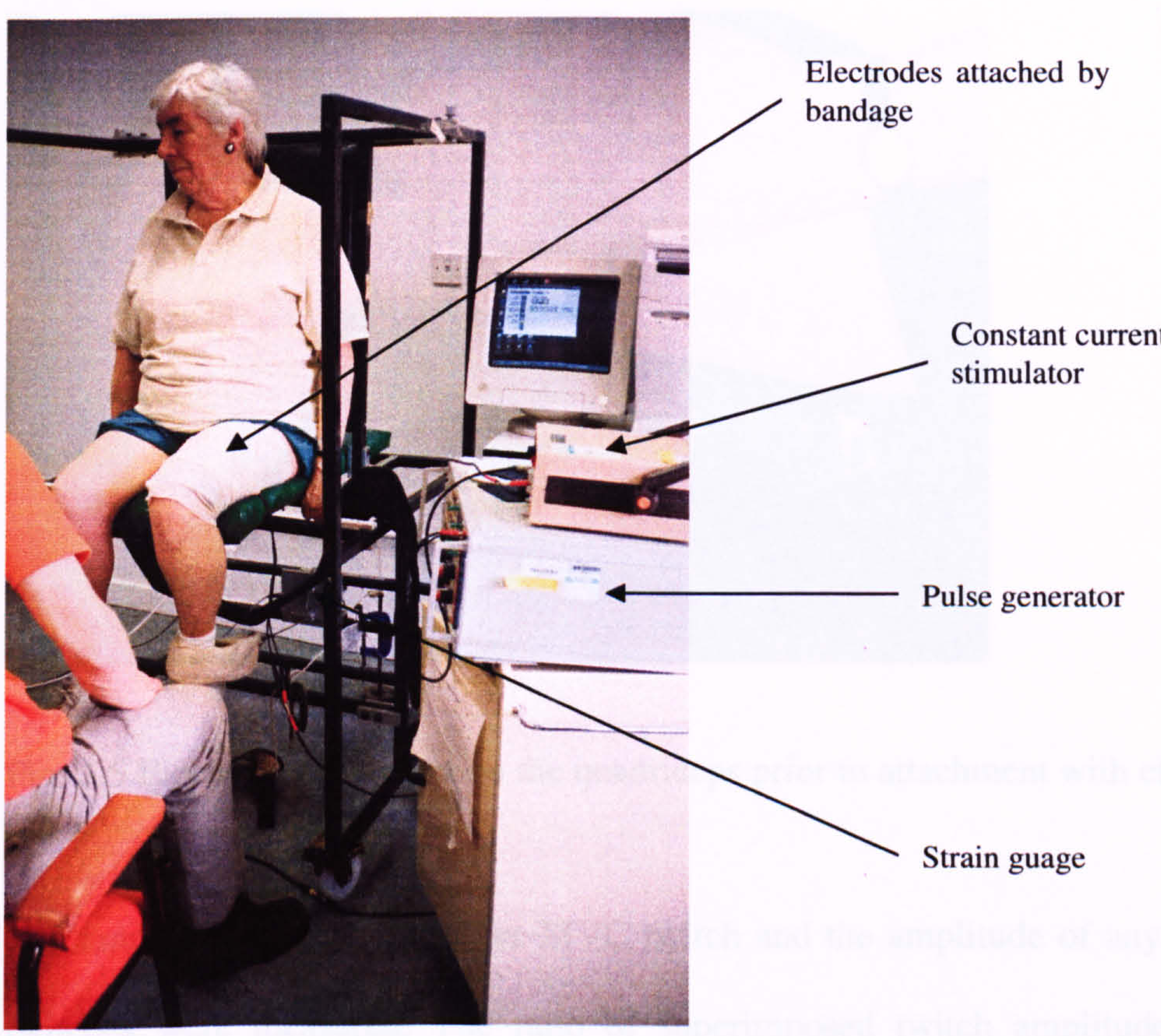


Fig. 3.4. Apparatus set-up for the measurement of quadriceps activation.

A constant current stimulator (DS7, Digitimer Ltd, UK) and programmable pulse generator (D4030, Digitimer Ltd, UK) were used to apply 200 μs 400mV square wave pulses at a frequency of 1Hz percutaneously to the quadriceps via 2 large (8cm x 12cm) rubber electrodes placed transversely over the proximal and distal areas of the quadriceps muscle (Fig. 3.5). Current was applied at the maximum intensity the subject could tolerate after familiarisation. After 4 to 5 muscle twitches had been obtained the subjects were asked to produce a maximum voluntary contraction (MVC) for 3 seconds while stimulation continued. This procedure was repeated twice, with a rest period of $\geq$ 1 min between each. Force data were fed into a purpose-built amplifier and the A-D converter (1401, Cambridge Electronic Devices (CED), UK). A PC was used to analyse force data using Signal 2.13 (CED, UK) software.

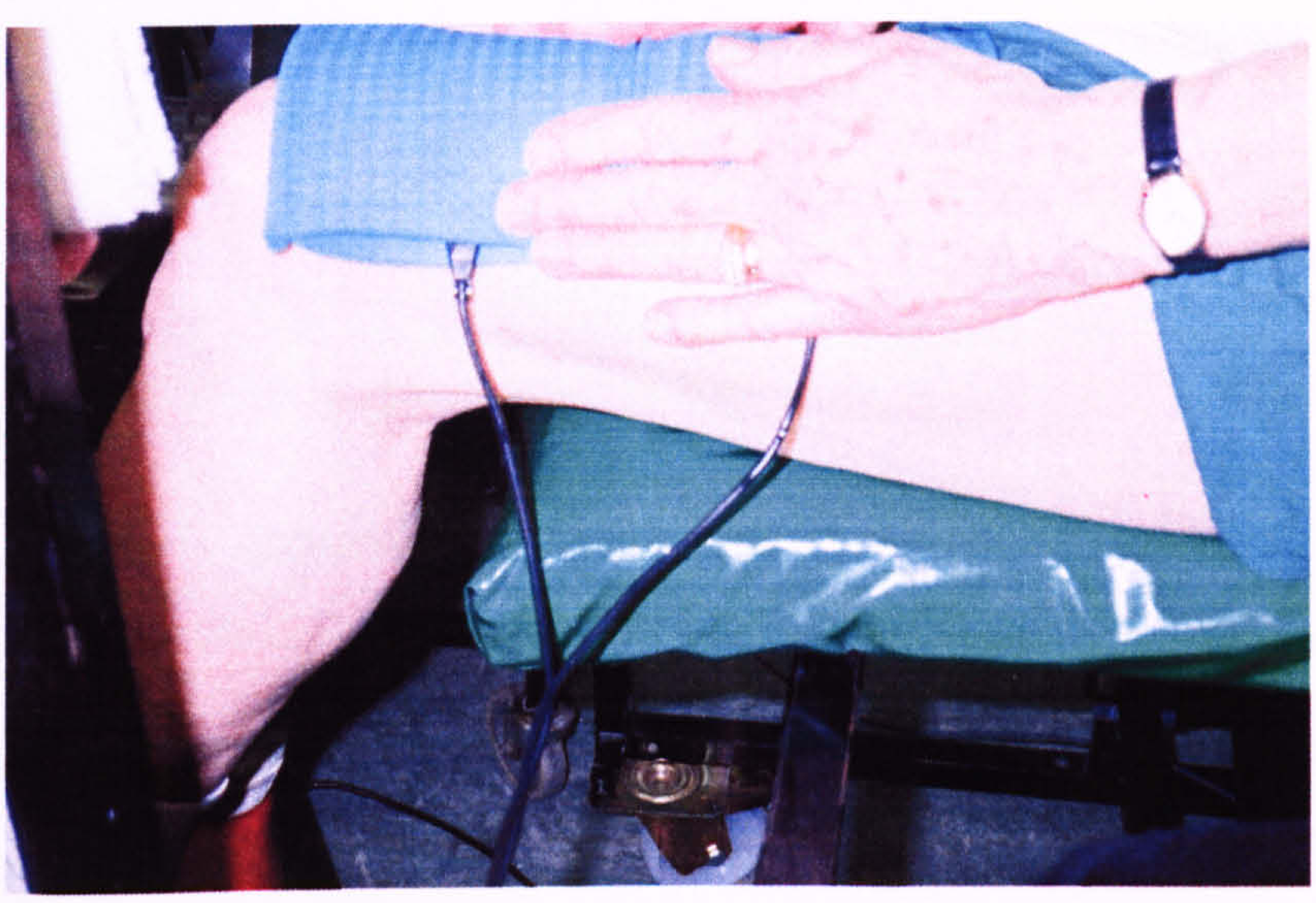


Fig. 3.5 Electrode placement on the quadriceps prior to attachment with elastic bandage

The amplitude of the largest pre-MVC twitch and the amplitude of any superimposed twitches were measured. The ratio of superimposed twitch amplitude to pre-MVC twitch amplitude was multiplied by 100. This was then subtracted from 100 to yield an estimate of percentage maximal voluntary activation of the quadriceps (Bigland-Richie et al. 1986, Allen et al. 1995). This was repeated for both legs. This technique of twitch interpolation has been shown to produce reliable results (Allen et al. 1995).

3.2.4 Statistical analysis

The analysis of all data in this chapter is described in chapter 2.

3.3 Results

3.3.1 Baseline group characteristics (Table 3.1)

The groups did not differ in body mass (P>0.05). The younger group were taller (P<0.001), but age-corrected height is more relevant to limb strength as it bears a more direct relationship with limb length (Sorkin et al. 2002). The square of height or limb length will be more related to strength than height or limb length alone (Astrand and Rodahl 1986) and so only age-corrected height squared was considered as a confounding variable. Age-corrected height squared did not differ between groups (P>0.05).

	Young		Older non-fallers		Post hoc p	Adjustments made for:
	n	Mean (SE)	n	Mean (SE)		
Weight (kg)	44	69.4 (1.61)	44	70.4 (1.60)	>0.05	sex
Height (m)	38	1.74 (0.01)	44	1.68 (0.01)	<0.01	sex
Age-corrected height (m)	38	1.72 (0.01)	44	1.73 (0.01)	>0.05	sex
Age-corrected height squared (m^2)	38	2.97 (0.04)	44	2.99 (0.04)	>0.05	sex
Male	18		15			
Female	26		29			

Table 3.1 Baseline characteristics of young and older subjects. The groups differed in sex ratios and height, but not in age-corrected height or age-corrected height squared

3.3.2 Interactions with sex

Sex interacted with all strength, CSA, power and strength/power variables (P<0.05), and male sex led to an increase in the value. For the only other variables interacting with sex, male sex led to an increase in the ratio of slow to fast hamstring concentric strength in the strong leg, and a decrease in the ratio of isometric to eccentric hamstring strength in the weak leg (P<0.05). Group comparisons were corrected for these interactions.

3.3.3 Strength

3.3.3.i Isometric and isokinetic strength

The young subjects had significantly higher isometric strength in all muscle groups in both legs (P<0.01). The angle-tension relationships appeared qualitatively similar in both groups, with peak forces occurring at the same angle. Only data for the stronger leg are shown as the weaker legs were qualitatively similar (Figs. 3.6-3.7).

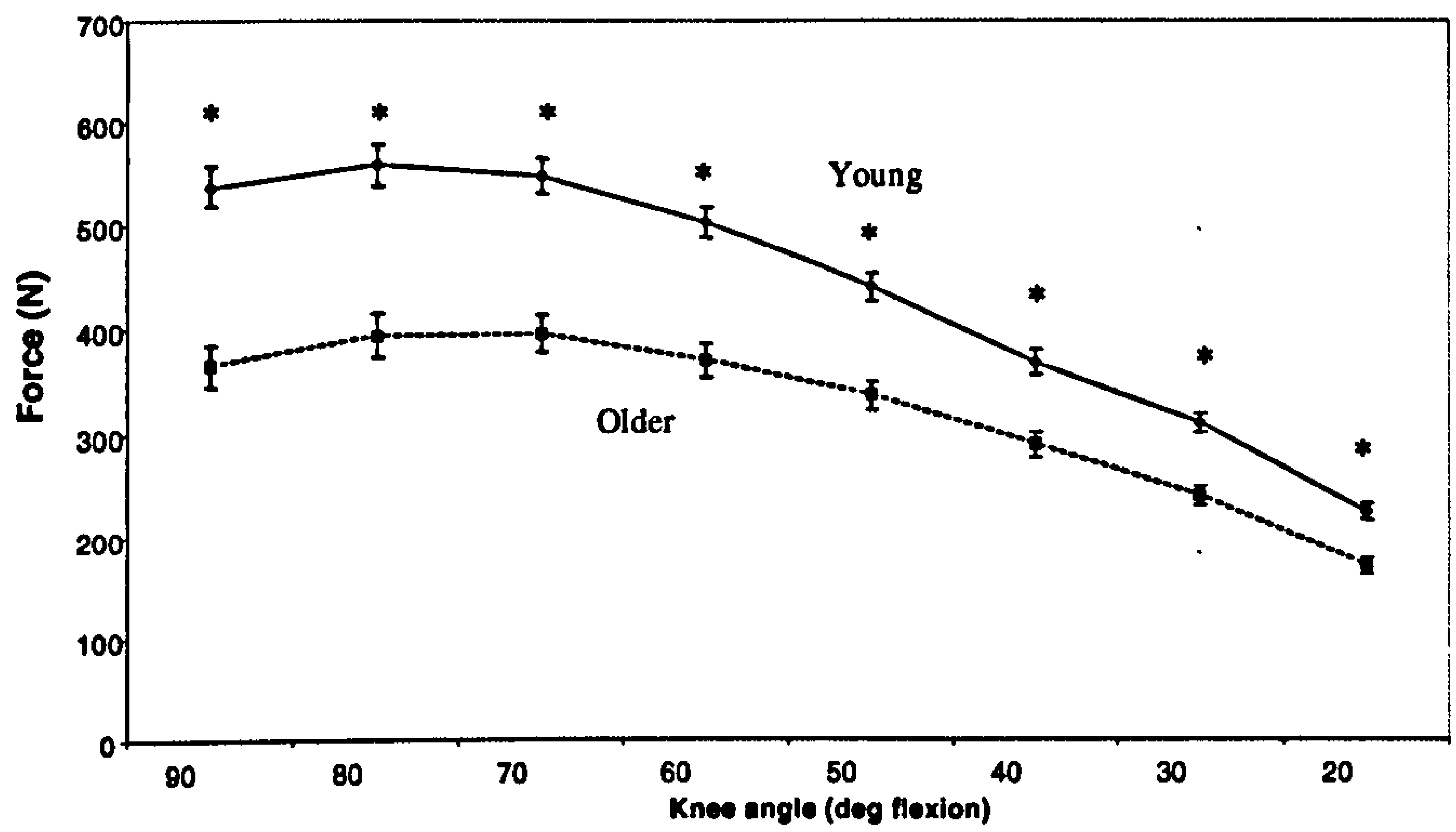


Fig. 3.6 Isometric quadriceps strength in young (n=37-41) and older (n=39-40) subjects. * P<0.01 (N = Newton)

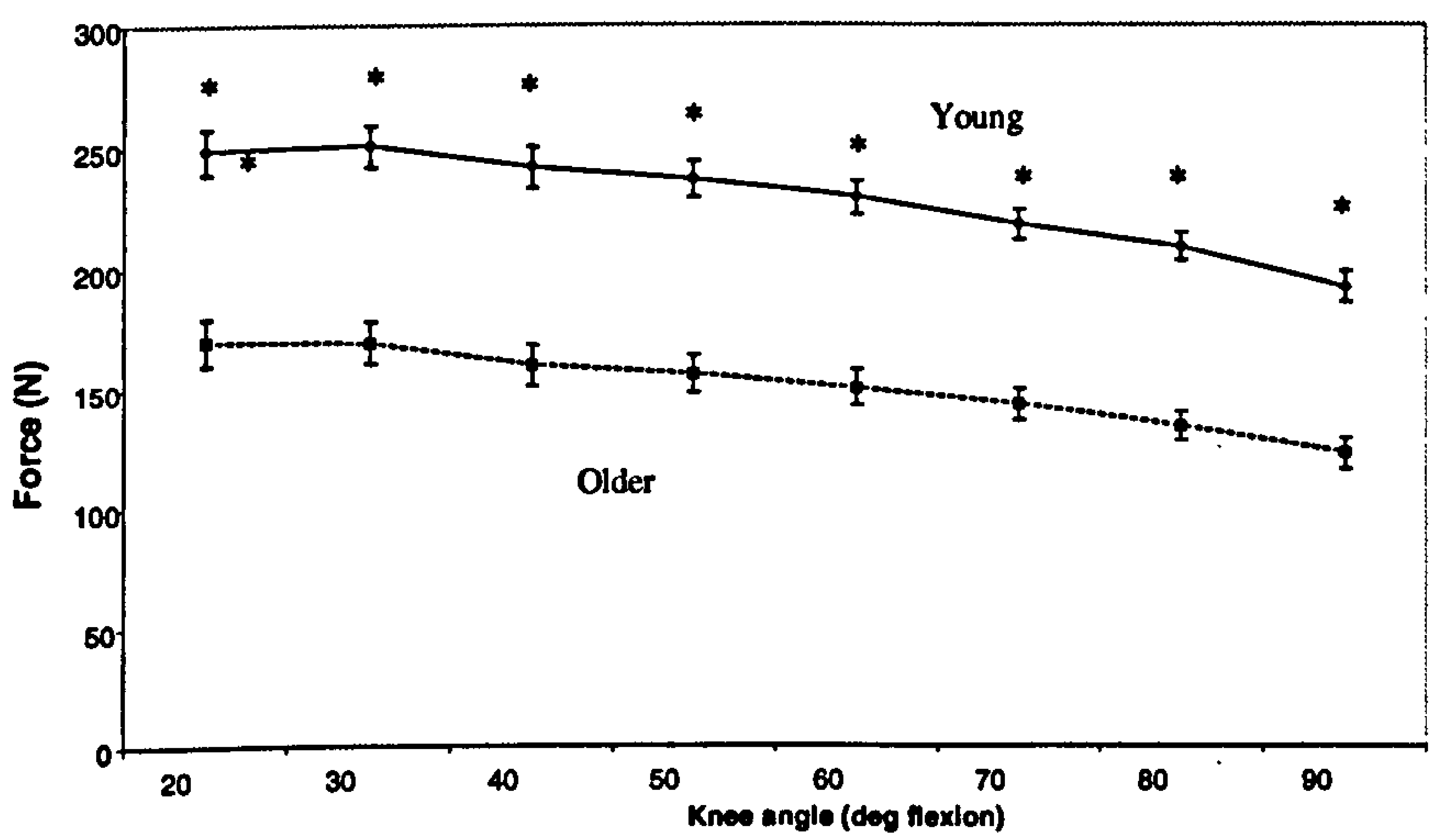


Fig. 3.7 Isometric hamstring strength in young (n=34-42) and older (n=38-40) subjects. * P<0.01 (N = Newton)

The young had significantly higher concentric and eccentric strength in both muscle groups at both speeds bilaterally. Only the stronger leg results are shown below as the weaker legs were qualitatively similar (Figs. 3.8-3.9).

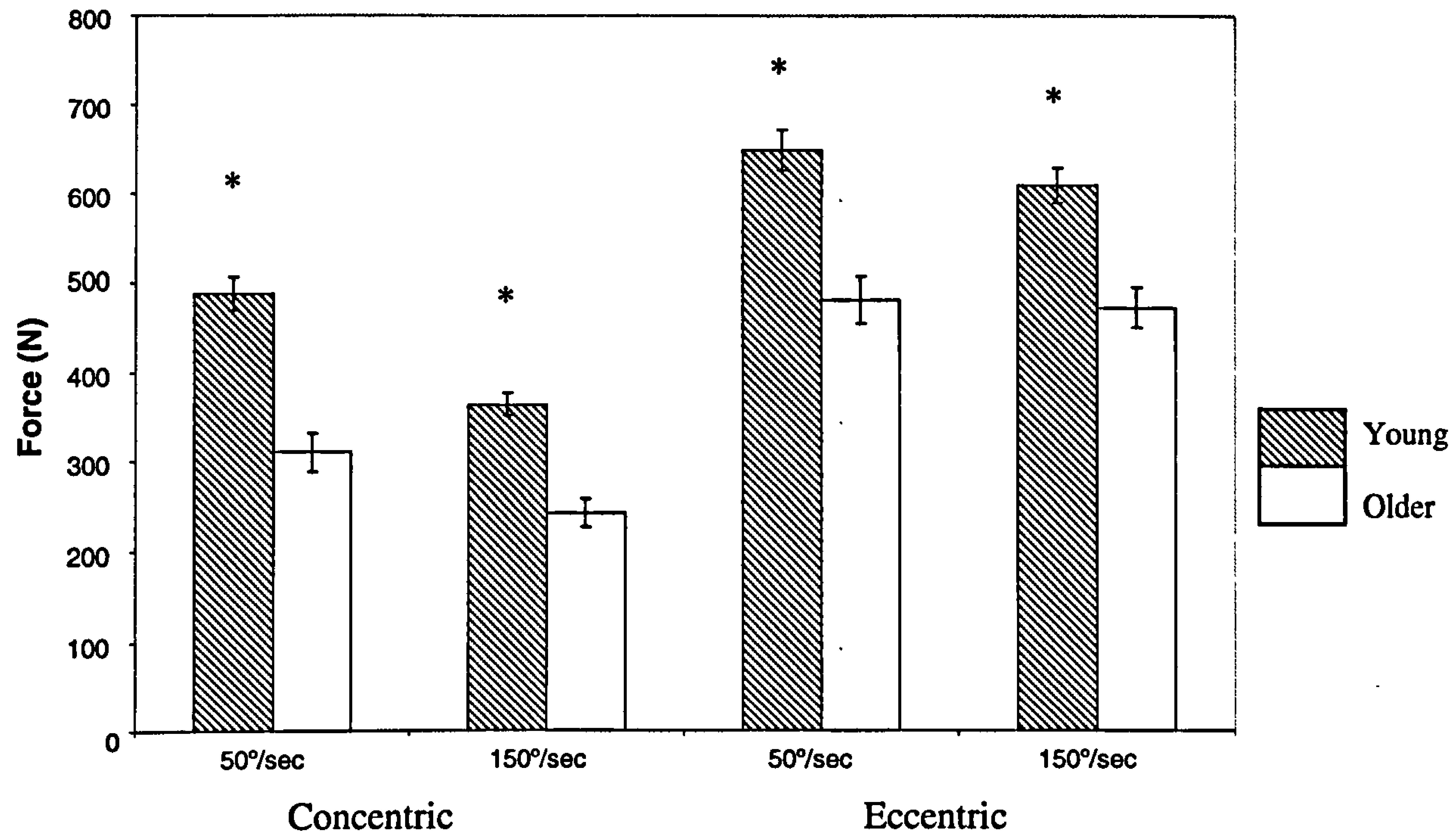


Fig. 3.8 Isokinetic quadriceps strength in young (N=38-39) and older (n=31-32) subjects. * P<0.01. (N = Newton)

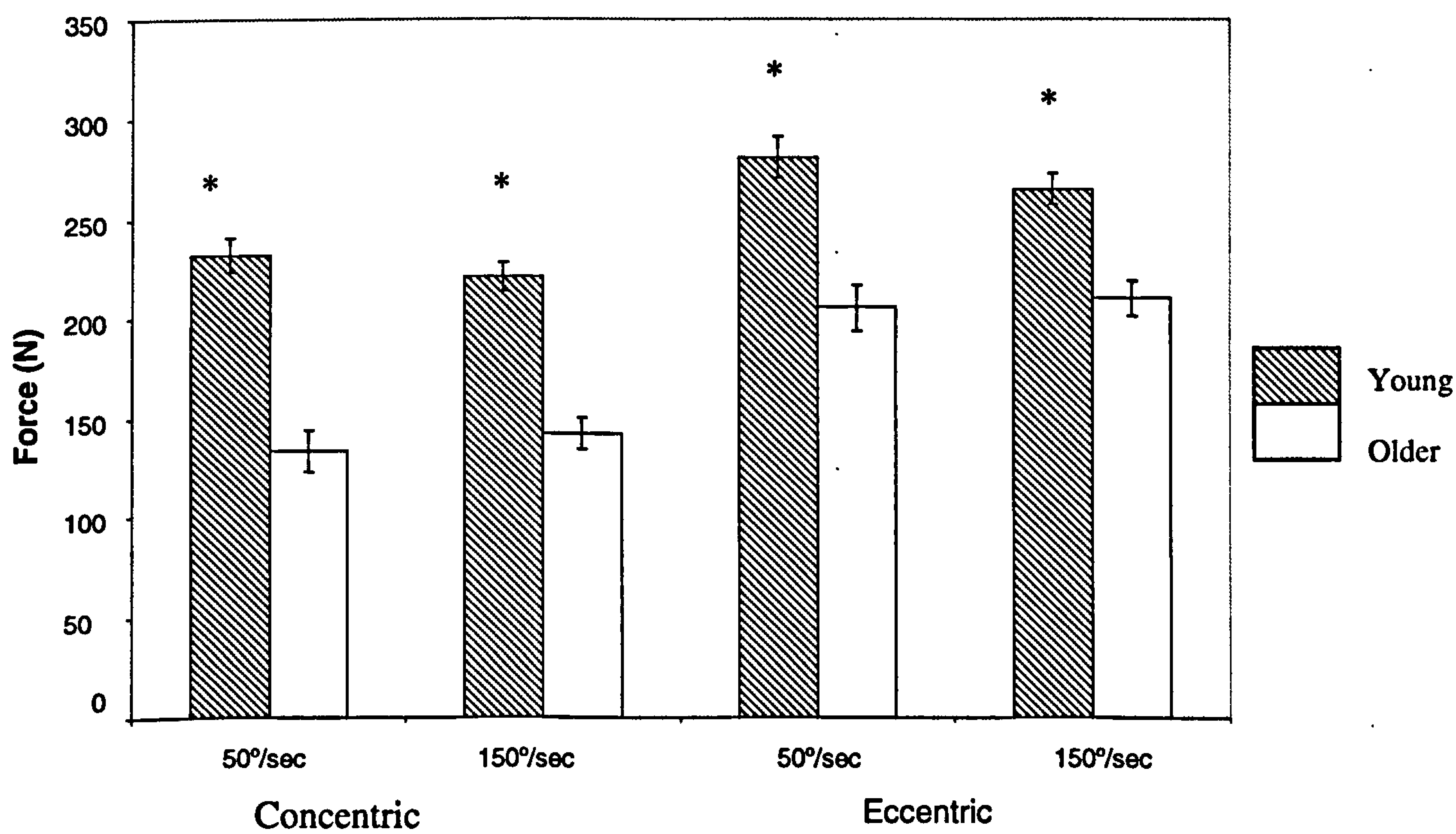


Fig. 3.9 Isokinetic hamstring strength in young (n=39-40) and older (n=32-34) subjects. * P<0.01. (N = Newton)

3.3.3.ii Normalised strength – body mass and corrected height squared

The strength values were normalised to body mass and also to corrected height squared to assess any subtle effects of these parameters. No changes in results were noted. It can therefore be assumed that body mass and corrected height squared did not significantly affect findings and so the discussion will focus on non-normalised results. Means and SE data are in Appendix 2.

3.3.3.iii Ratios of different contraction types and speeds

Ratios of forces in the different contraction types, and also across different speeds within the same contraction type, were calculated to assess the relative changes of forces in contraction types or speeds with age. When comparing contraction types, the contractions at $50°.sec^{-1}$ only were used. Averaging values across muscles and legs revealed the young had significantly higher isometric to eccentric (young 0.92 ± 0.03, older 0.83 ± 0.03, $P<0.05$), concentric to eccentric (young 0.81 ± 0.02, older 0.61 ± 0.02, $P<0.01$), and slow to fast eccentric force ratios (young 1.05 ± 0.02, older 0.97 ± 0.02, $P<0.01$), indicating that eccentric contractions declined less with age than concentric or isometric contractions, and that eccentric forces declined less with age at higher velocities. The young also had significantly lower isometric to concentric force ratios (young 1.15 ± 0.03, older 1.45 ± 0.06, $P<0.01$), indicating that isometric contractions declined less with age than concentric contractions. The averaged slower to faster concentric force ratios did not differ, suggesting no overall differences in the rates of fast and slow concentric decline.

When variables were considered individually, in the quadriceps the young group had a significantly higher concentric to eccentric ratio bilaterally, and a significantly lower isometric to concentric force ratio in the weaker leg. In the hamstrings the young had a significantly lower isometric to concentric force ratio bilaterally, and significantly higher concentric to eccentric, isometric to eccentric and slow to fast concentric force ratios bilaterally. The young also had a significantly higher slower to faster eccentric force ratio in the stronger leg. There were no further differences between groups. Figs. 3.10-3.13 summarise these findings.

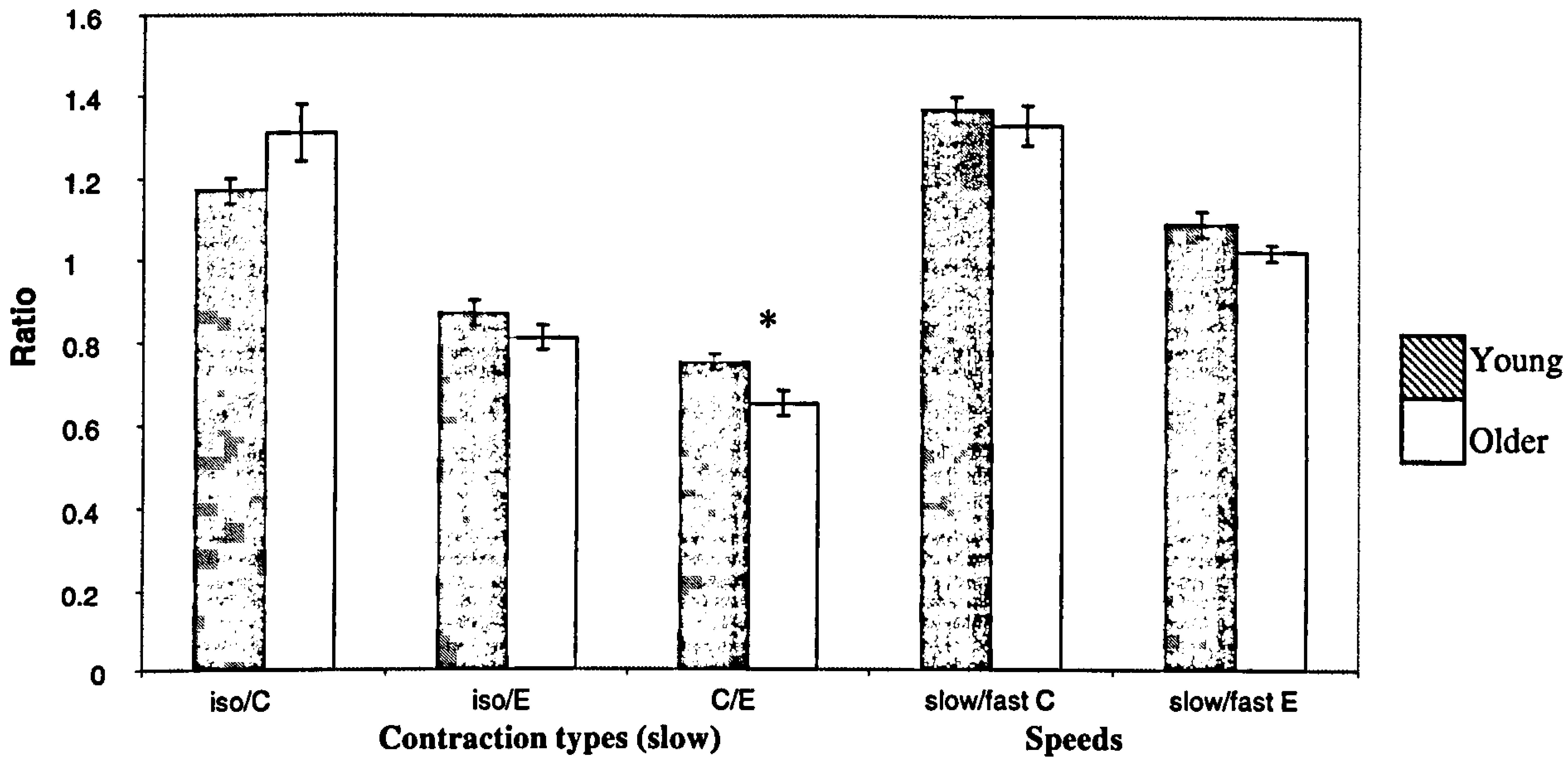


Fig. 3.10 Quadriceps strength ratios for contraction types and speeds in young and older subjects in the stronger leg. * P<0.01. iso=isometric, C=concentric, E=eccentric, slow= 50°.sec^{-1} fast=150°.sec^{-1}

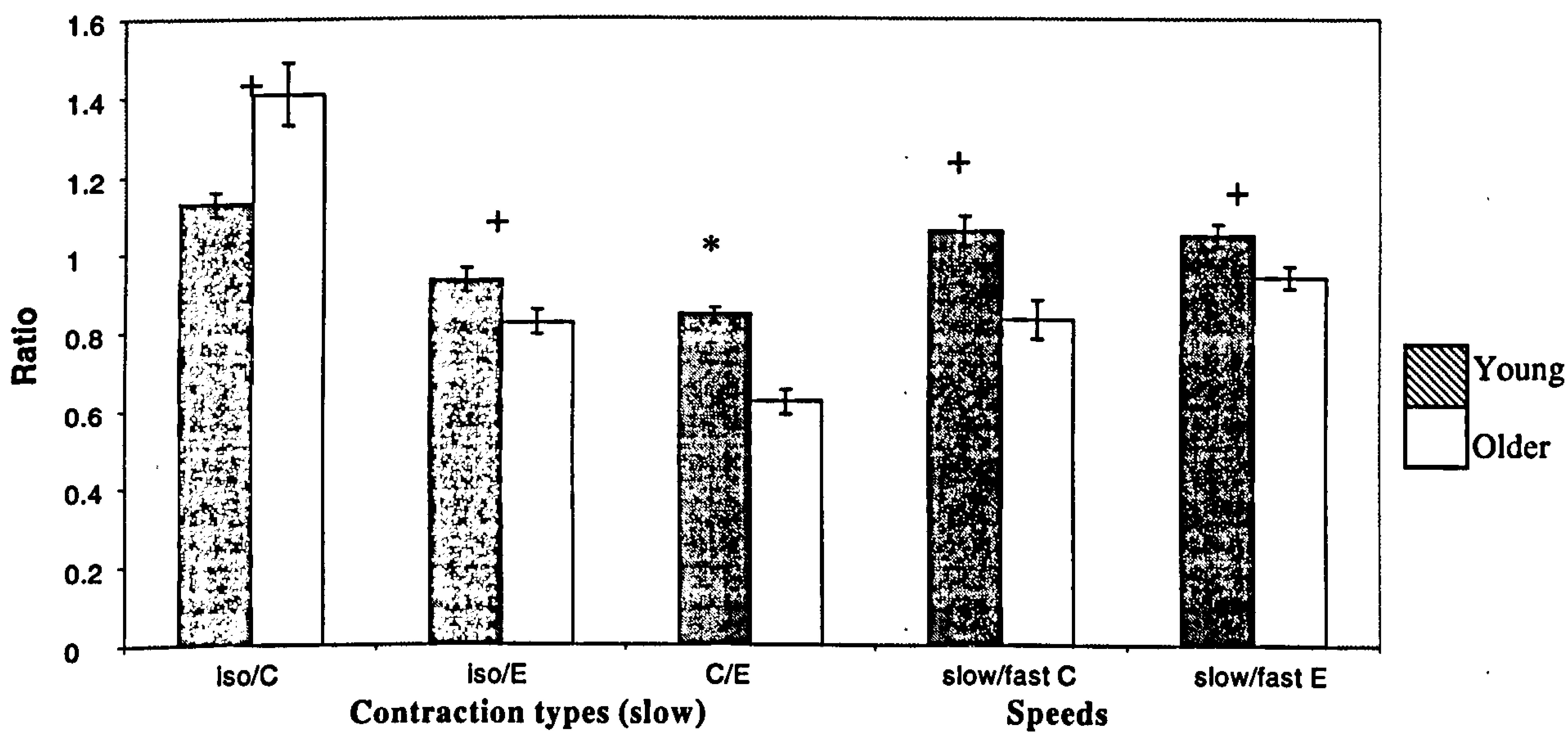


Fig. 3.11 Hamstrings strength ratios for contraction types and speeds in young and older non-fallers in the stronger leg. + P<0.05, *P<0.01. iso=isometric, C=concentric, E=eccentric, slow=50°.sec^{-1} fast=150°.sec^{-1}

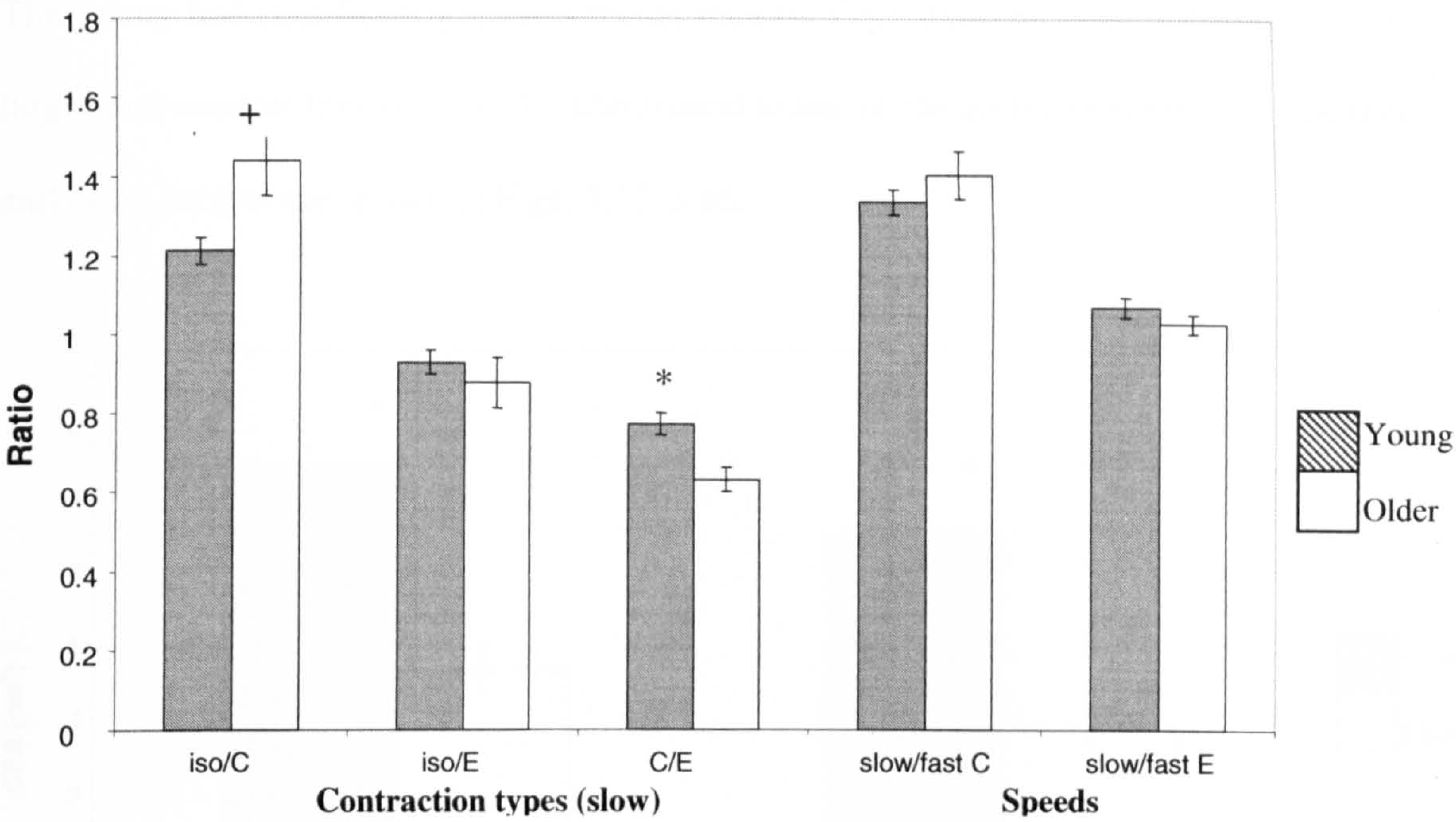


Fig. 3.12 Quadriceps strength ratios for contraction types and speeds in young and older subjects in the weaker leg. + $P<0.05$ * $P<0.01$. iso=isometric, C=concentric, E=eccentric, slow=50°.sec^{-1} fast=150°.sec^{-1}

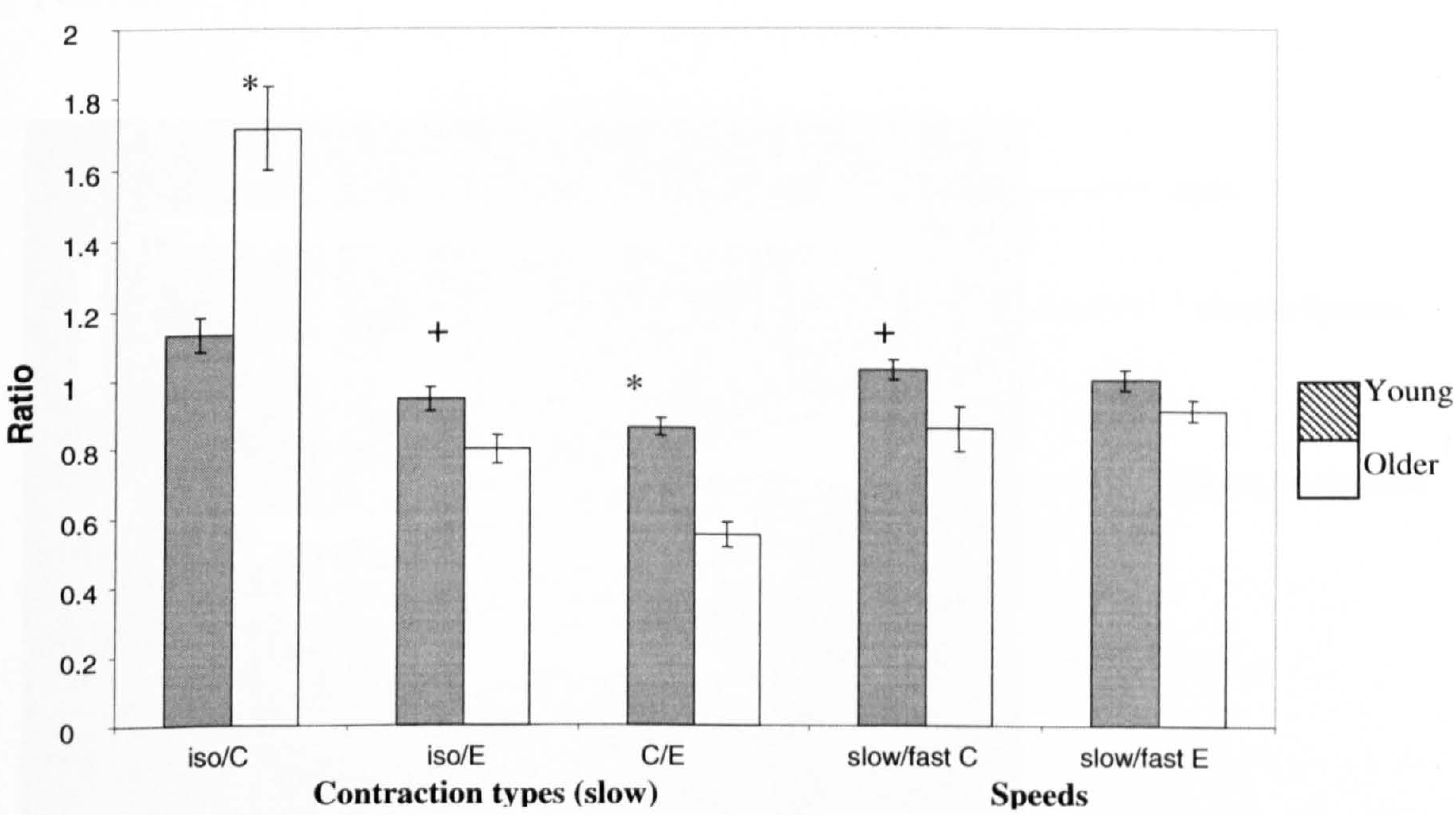


Fig. 3.13 Hamstrings strength ratios for contraction types and speeds in young and older subjects in the weaker leg. + $P<0.05$ * $P<0.01$. iso=isometric, C=concentric, E=eccentric, slow=50°.sec^{-1} fast=150°.sec^{-1}

3.3.4 Age effects on rectus femoris CSA

The young had significantly greater rectus femoris CSA than the older subjects in both larger and smaller legs (Fig. 3.14). Ultrasound scans of the rectus femoris of a younger and older subject are shown in Figs. 3.15-3.16.

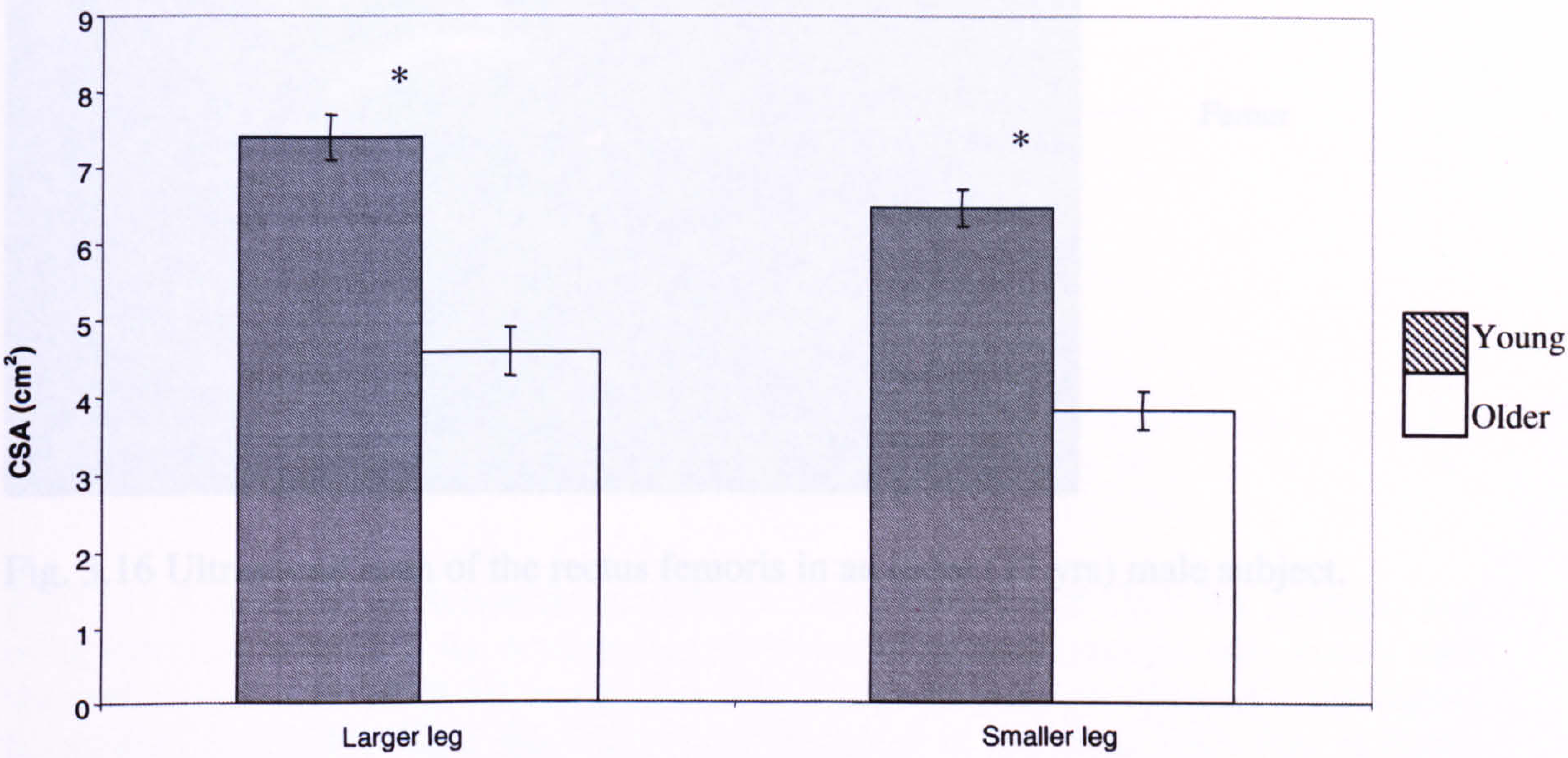


Fig. 3.14 Rectus femoris CSA in young (n=31) and older (n=29) subjects in both legs * P<0.001.

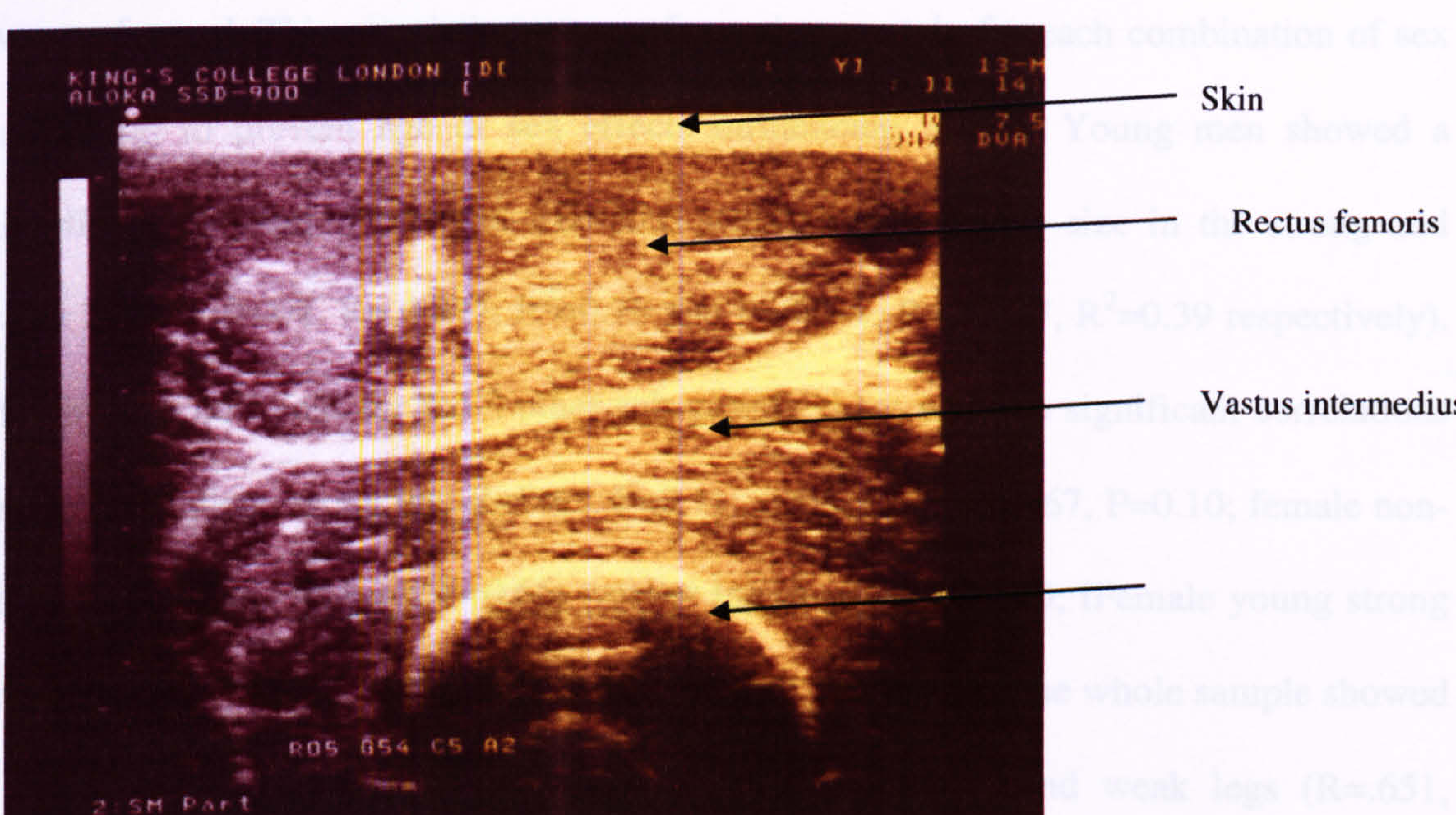


Fig. 3.15 Ultrasound scan of the rectus femoris in a young (37 yrs) male subject.

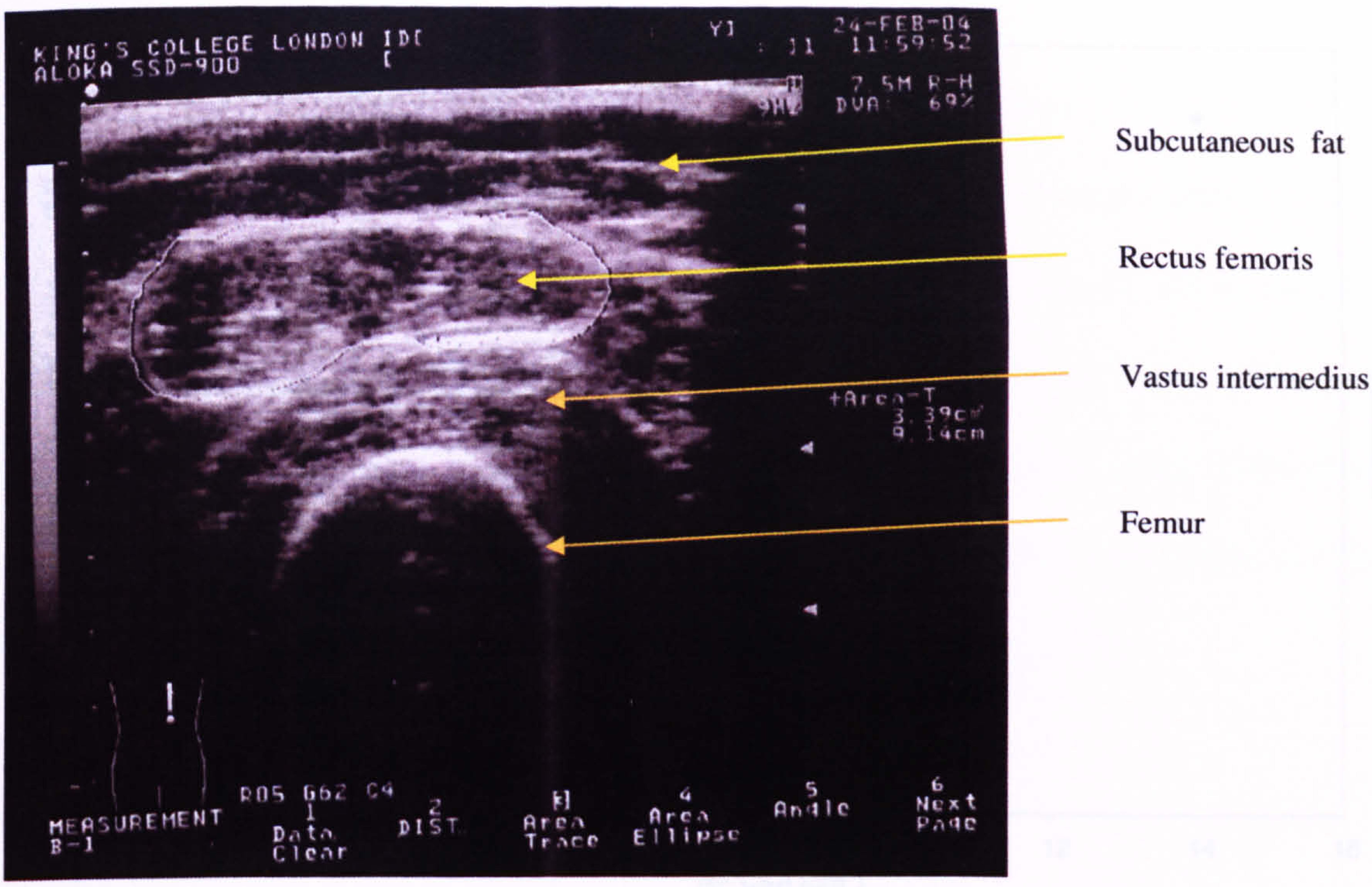


Fig. 3.16 Ultrasound scan of the rectus femoris in an older (77 yrs) male subject.

A correlation analysis between the rectus femoris area (in the stronger and weaker legs) and quadriceps isometric strength at 80° knee flexion (in the stronger and weaker legs) was performed. This correlation was performed separately for each combination of sex and group to prevent age or sex effects influencing results. Young men showed a significant correlation between isometric strength and muscle size in the strong and weak legs (P=0.001, R= 0.837, R^2=0.70 and P=0.039, R= 0.627, R^2=0.39 respectively). However in both legs for the 3 other sub-groups, there were no significant correlations (male non-fallers strong leg: R=0.61, P=0.11; weaker leg: R=0.67, P=0.10; female non-fallers strong leg: R=0.11, P=0.65; weaker leg: R=0.21, P=0.40; fFemale young strong leg: R=-0.44, P=0.89; weaker leg: R=-0.402, P=0.12). Overall the whole sample showed a significant correlation in the strong (R=.737, P<0.001) and weak legs (R=.651, P<0.001, Figs. 3.17-3.18).

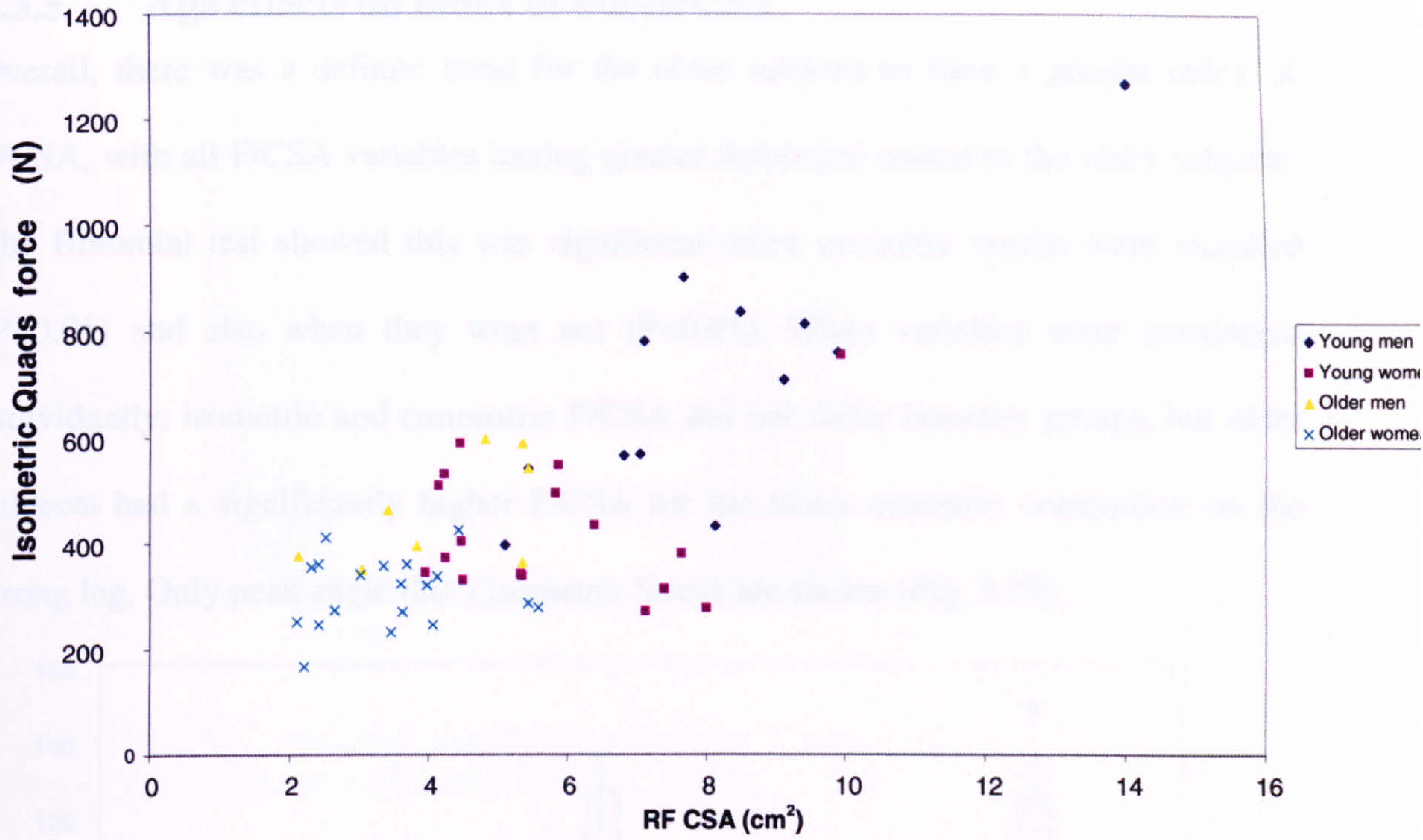


Fig. 3.17 Scatterplot of isometric quadriceps force against RF CSA for younger and older men and women in the stronger leg. In terms of separate groups, only young men showed a significant (P<0.01) correlation. The whole sample also showed a significant correlation (P<0.001). Note the wide range in forces at the same CSA (i.e. at 8cm^2, forces range from ~300 to ~900N). This indicates that individual differences in specific force have a high influence on individual variations in strength.

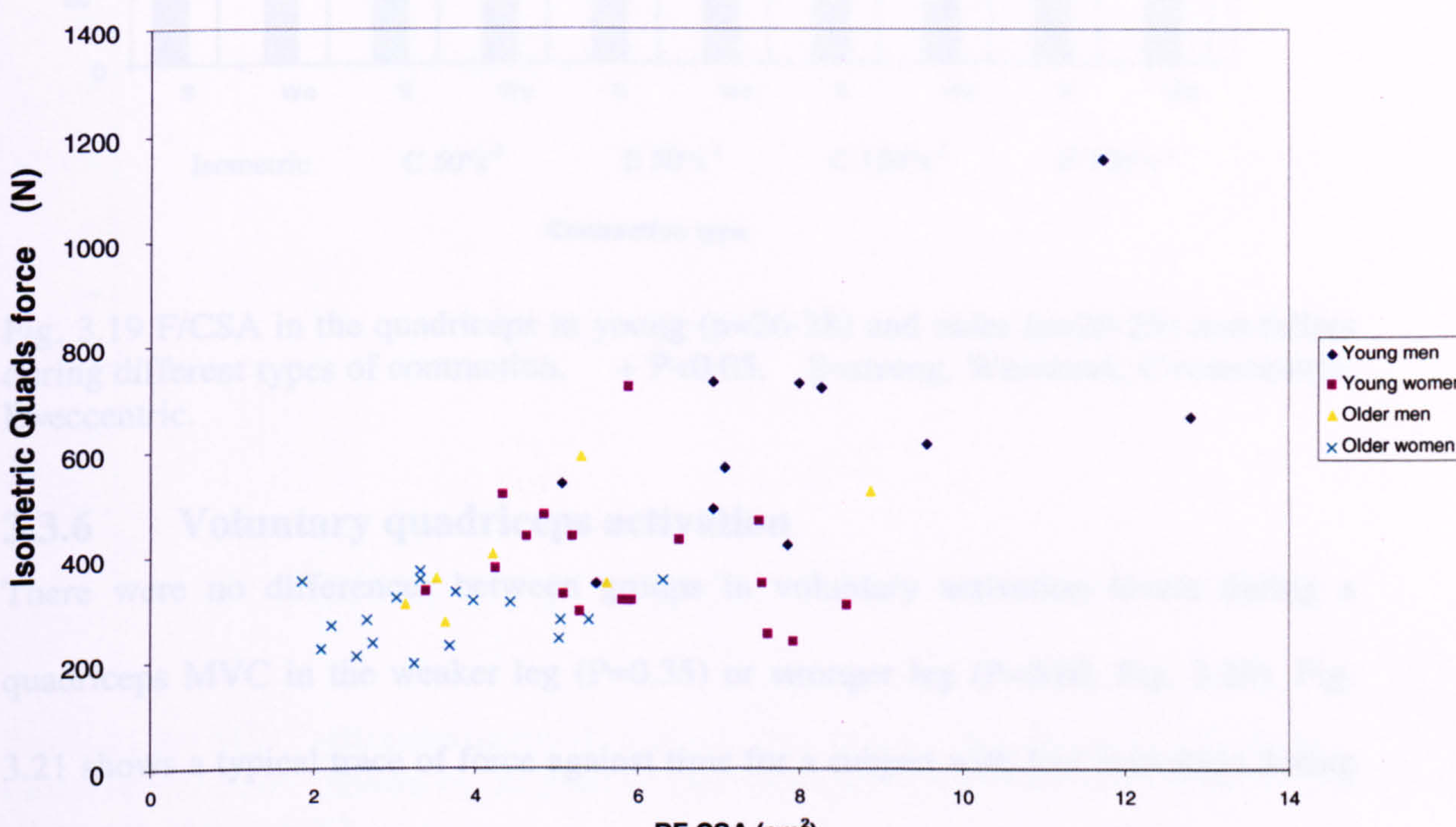


Fig. 3.18 Scatterplot of isometric quadriceps force against RF CSA for younger and older men and women in the weaker leg. Only the young men showed a significant correlation (P<0.05). The whole sample showed a significant correlation (P<0.001)

3.3.5 Age effects on index of Force/CSA

Overall, there was a definite trend for the older subjects to have a greater index of F/CSA, with all F/CSA variables having greater numerical means in the older subjects. The Binomial test showed this was significant when eccentric results were included ($P<0.01$) and also when they were not ($P<0.05$). When variables were considered individually, isometric and concentric F/CSA did not differ between groups, but older subjects had a significantly higher F/CSA for the faster eccentric contraction on the strong leg. Only peak angle (80°) isometric forces are shown (Fig. 3.19).

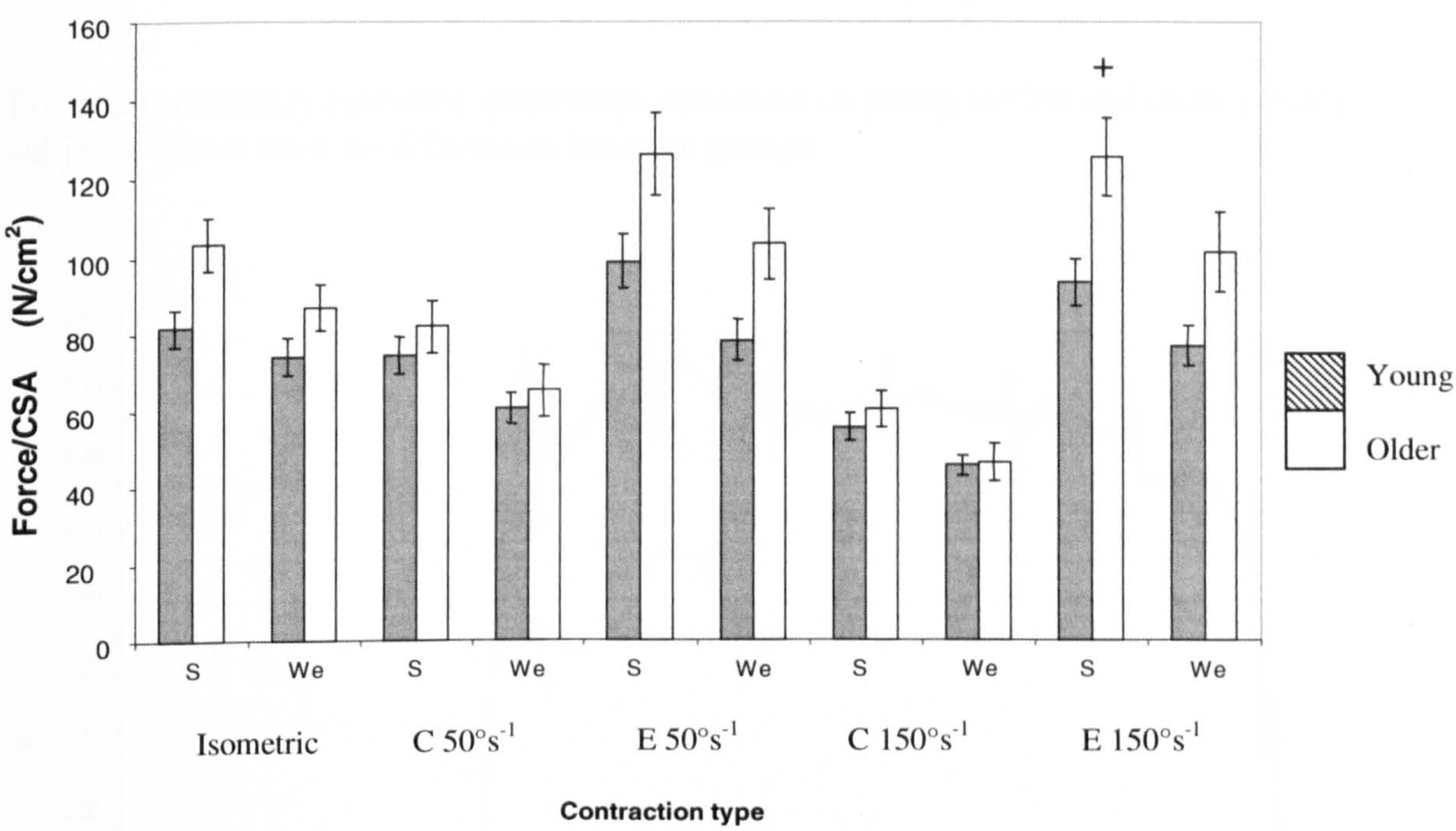


Fig. 3.19 F/CSA in the quadriceps in young (n=26-28) and older (n=20-25) non-fallers during different types of contraction. + $P<0.05$. S=strong, We=weak, C=concentric, E=eccentric.

3.3.6 Voluntary quadriceps activation

There were no differences between groups in voluntary activation levels during a quadriceps MVC in the weaker leg ($P=0.35$) or stronger leg ($P=0.80$, Fig. 3.20). Fig. 3.21 shows a typical trace of force against time for a subject with low activation during this test.

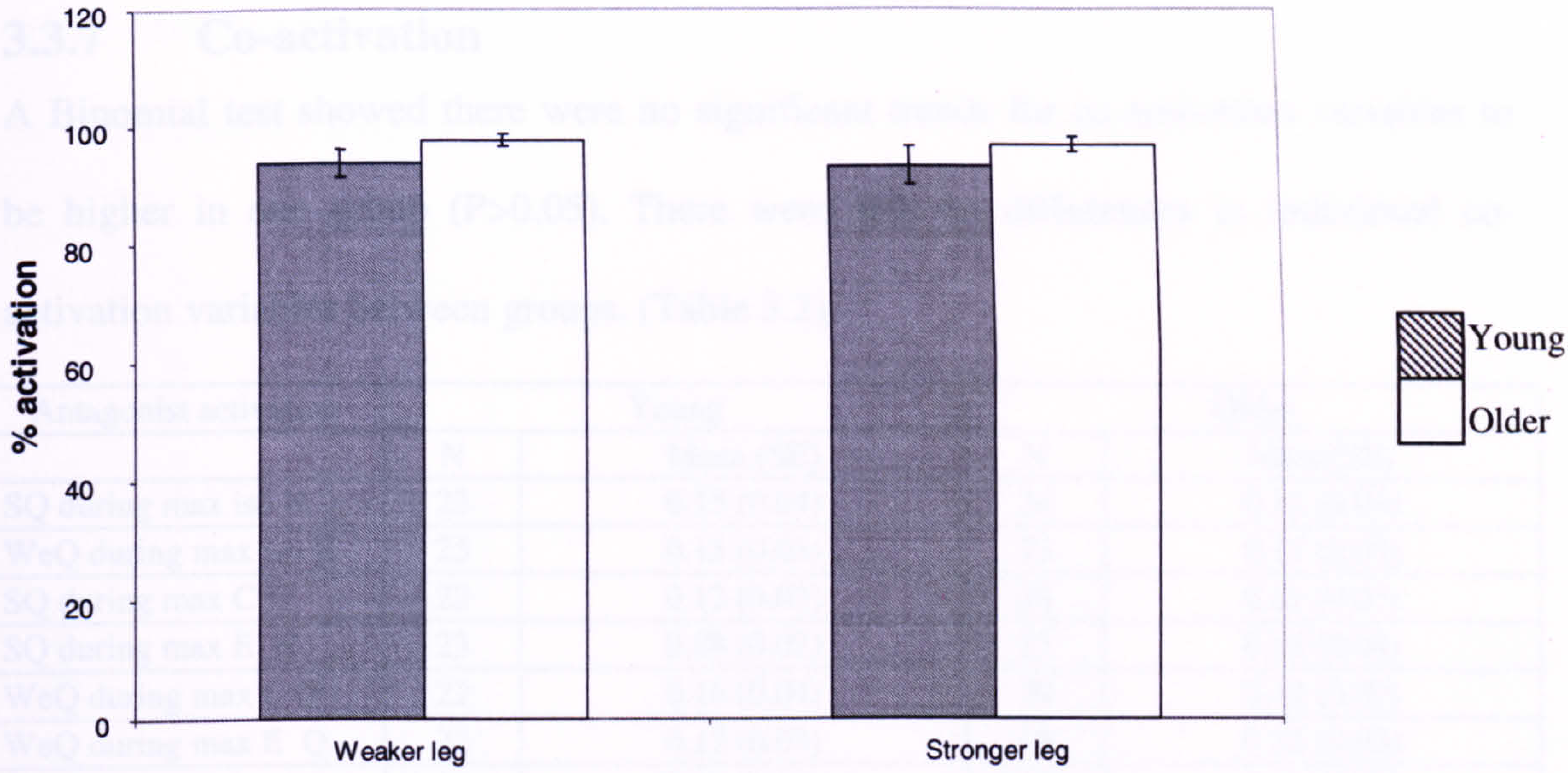


Fig. 3.20 Voluntary isometric quadriceps activation in young (n=21) and older (n=31) subjects. There were no differences between groups.

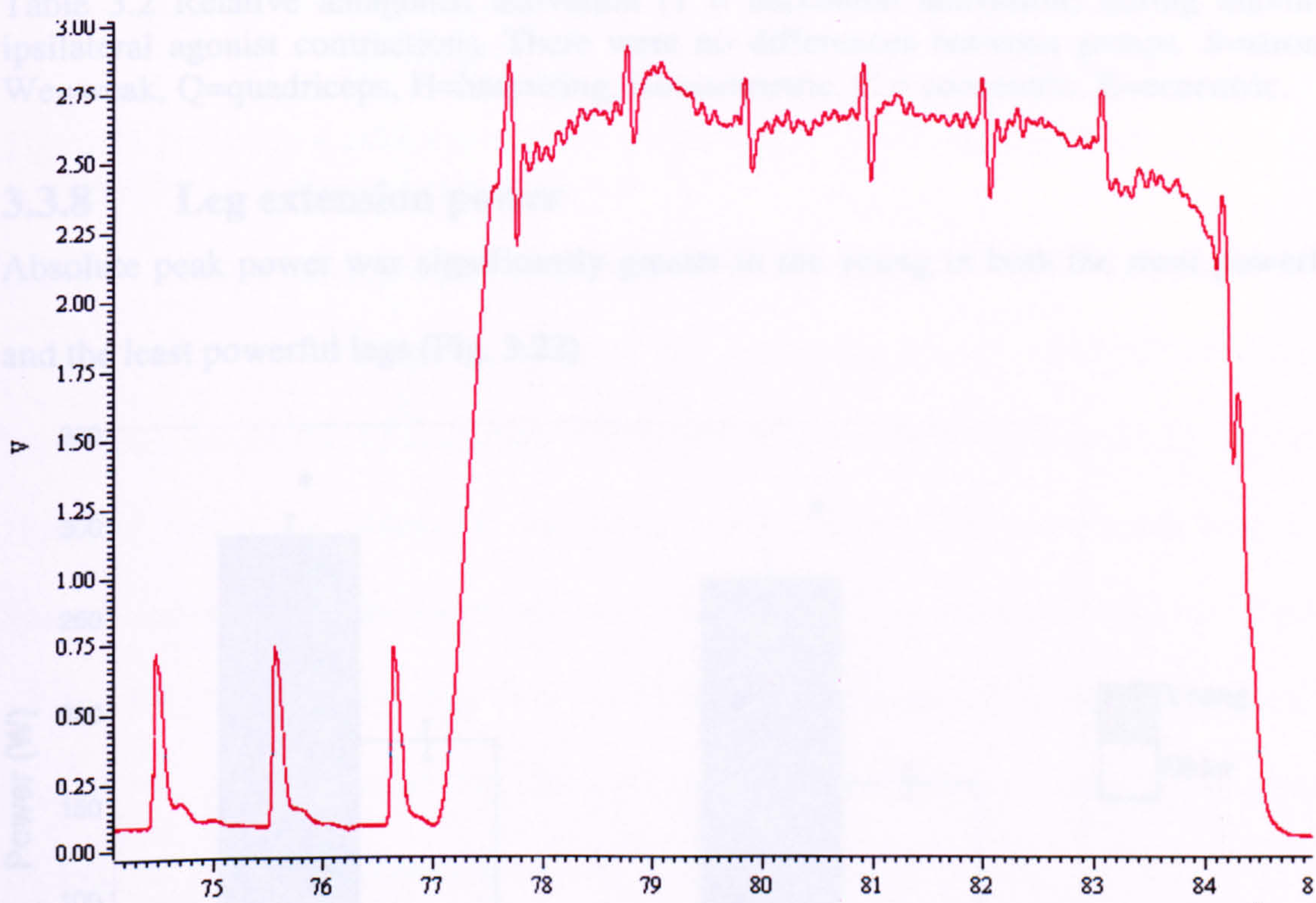


Fig. 3.21 An MVC with superimposed twitches. This subject had activated 71% of the motor neurone pool, shown by the positive sweeps of superimposed twitches being 29% of the pre-MVC twitch.

3.3.7 Co-activation

A Binomial test showed there were no significant trends for co-activation variables to be higher in one group (P>0.05). There were also no differences in individual co-activation variables between groups. (Table 3.2).

Antagonist activation	Young		Older	
	N	Mean (SE)	N	Mean(SE)
SQ during max iso H	23	0.15 (0.04)	24	0.12 (0.03)
WeQ during max iso H	25	0.15 (0.03)	23	0.17 (0.05)
SQ during max C H	22	0.12 (0.03)	24	0.21 (0.07)
SQ during max E H	23	0.08 (0.02)	25	0.15 (0.04)
WeQ during max C Q	22	0.16 (0.04)	20	0.12 (0.02)
WeQ during max E Q	22	0.12 (0.03)	18	0.12 (0.03)
SH during max iso Q	22	0.21 (0.04)	22	0.14 (0.02)
WeH during max iso H	20	0.22 (0.05)	19	0.24 (0.05)
SH during max C Q	18	0.17 (0.04)	19	0.22 (0.04)
SH during max E Q	19	0.15 (0.03)	18	0.17 (0.04)
WeH during max C H	18	0.24 (0.05)	18	0.21 (0.04)
WeH during max E H	17	0.22 (0.05)	17	0.28 (0.06)

Table 3.2 Relative antagonist activation (1 = maximum activation) during maximal ipsilateral agonist contractions. There were no differences between groups. S=strong, We=weak, Q=quadriceps, H=hamstring, iso=isometric, C = concentric, E=eccentric.

3.3.8 Leg extension power

Absolute peak power was significantly greater in the young in both the most powerful and the least powerful legs (Fig. 3.22)

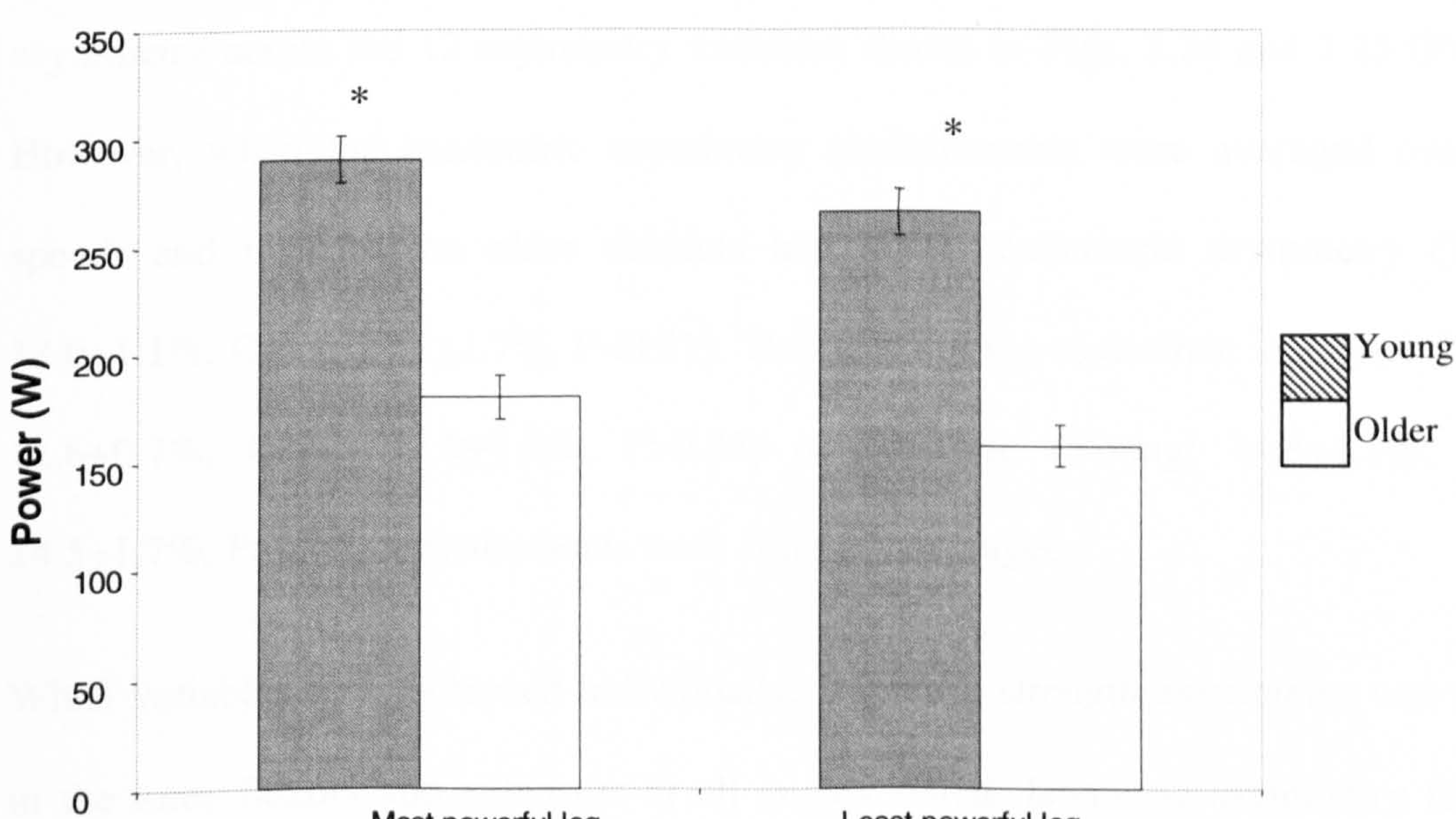


Fig. 3.22 Maximum lower limb extensor power in young (n=38) and older (n=43) subjects. * P<0.01.

When isometric quadriceps strength was divided by power there were no differences between groups(Fig.3.23).

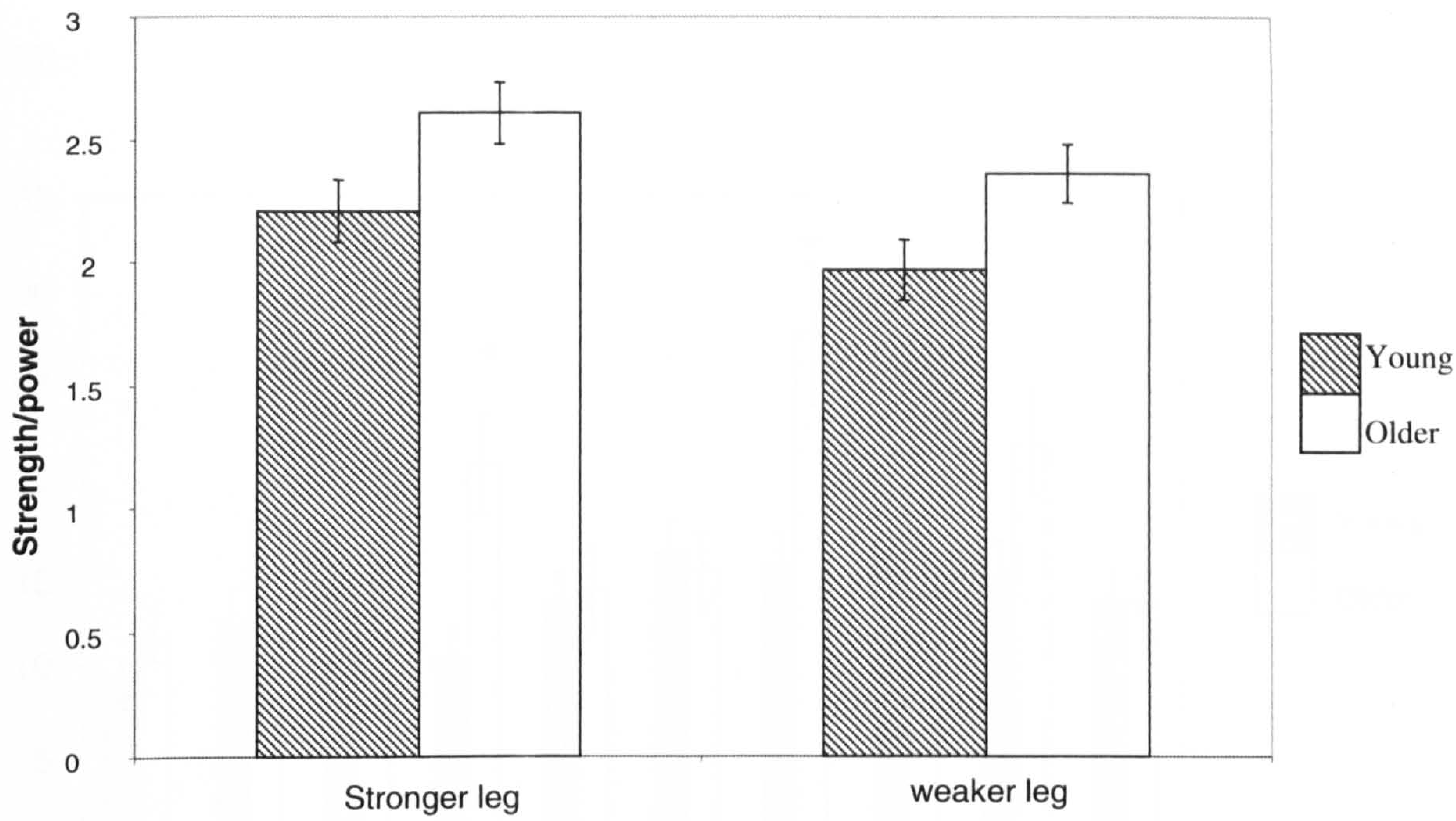


Fig. 3.23 Isometric quadriceps strength/power in young and older subjects. There were no significant differences between groups.

3.3.9 Asymmetry of strength and power

Overall, there was no significant trend for one group to have numerically greater asymmetry across the 12 asymmetry variables shown in Figs. 3.24 and 3.25 ($P>0.05$). However, when the concentric asymmetry measurements were averaged over both speeds and muscles the older subjects had higher concentric asymmetry (Young: 13.0$\pm$1.1%, Older: 21.7$\pm$1.7%, $P<0.01$). This was not the case when isometric (Young: 12.6$\pm$0.7%, Older: 13.9$\pm$1.0%, $P>0.05$) or eccentric (Young: 14.3$\pm$1.1%, Older: 14.5$\pm$1.7%, $P>0.05$) measurements were similarly averaged.

When variables were anlaysed individually, isometric strength asymmetry was similar in the knee flexors and extensors at all angles tested. Isometric asymmetry for each muscle group at the angle producing the highest force is shown (Fig. 3.24). There were also no differences in eccentric asymmetry but for concentric contractions the older group had greater asymmetry in the quadriceps contraction at 150°.sec^{-1} and the hamstring contraction at 50°.sec^{-1} (Fig. 3.24). There were no group differences in rectus femoris cross-sectional area asymmetry or lower limb extensor power asymmetry (Fig. 3.25).

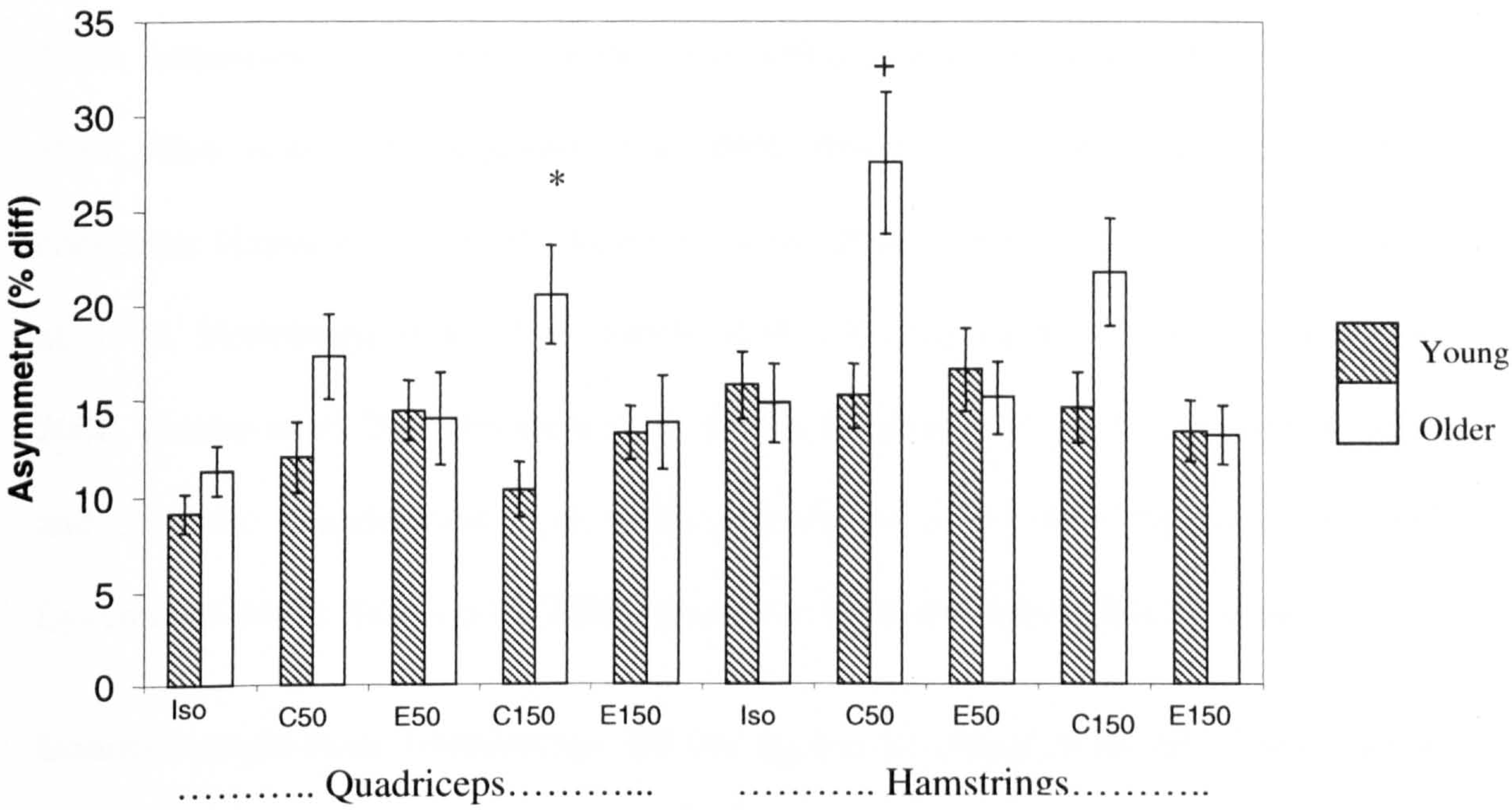


Fig. 3.24. Asymmetry of muscle strength in young (n=32-40) and older (n=32-34) people + P<0.05 *P<0.01. Q=quads, H=Hamstrings, C=concentric, E=eccentric, Iso = peak isometric, number represents angular velocity in °.sec^{-1}

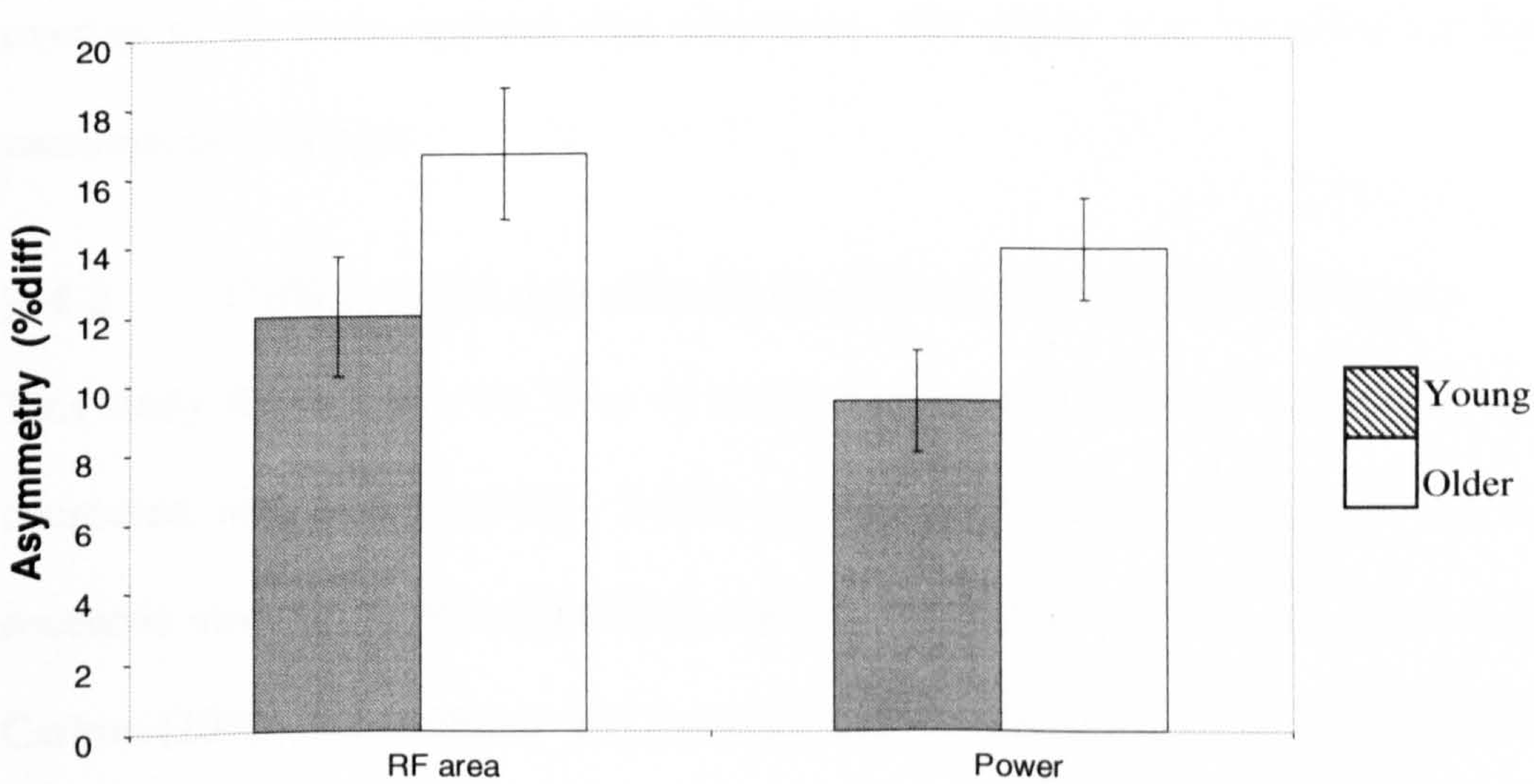


Fig. 3.25. Asymmetry of power and rectus femoris CSA in young (n = 31-36) and older (n = 29-43) people. There were no differences between groups.

3.4 Discussion

3.4.1 Age effects on strength

The results from this study confirmed those of previous studies into age-related declines in isometric (Larsson et al. 1979, Aniansson et al. 1983, Young et al. 1984, Young et al. 1985, Aniansson et al. 1986, Overend et al. 1992, Hortobagyi et al. 1995, Lindle et al. 1997, Roos et al. 1999, Izquierdo et al. 1999, Hunter et al. 2000, Lanza et al. 2003), concentric (Larsson et al. 1979, Aniansson et al. 1986, Frontera et al. 1991, Overend et al. 1992, Hortobagyi et al. 1995, Lindle et al. 1997, Lynch et al. 1999, Akima et al. 2000, Hunter et al. 2000, Frontera et al. 2000a, Hughes et al. 2001, Lanza et al. 2003) and eccentric (Vandervoort et al. 1990, Hortobagyi et al. 1995, Lindle et al. 1997, Lynch et al. 1999, Akima et al. 2000) strength in the quadriceps and hamstrings.

Isometric angle-force relationships did not appear to change with age. These can be regarded as an index of length-tension relationships in single fibres. Narici et al. (2003) have argued that although decreased tendon stiffness with age might increase force at longer muscle lengths, due to the increased tendon compliance allowing sarcomere overlap to be more optimal than otherwise, this effect may be offset by the loss of sarcomeres with age.

3.4.2 Differential age effects in different contraction types

This study showed that the force of isometric, concentric and eccentric contractions all decreased with age. Similarly, Lindle et al. (1997) reported that both concentric and eccentric strength in the quadriceps significantly declined with age, and Christou and Carlton (2002) and Delbaere and Bourgois (2003) reported similar declines in isometric, concentric and eccentric quadriceps strength with age. In contrast, some studies have shown no changes in quadriceps eccentric strength with age in the presence of decreases in other types of contraction (Hortobagyi et al. 1985, Poulin et al. 1992).

Although the magnitude of muscle force was greater in younger subjects for all contraction types, there were differences in the degree to which the different contraction types changed with age. The method of comparing young and old for the ratios of different contraction types was employed to estimate how the different contraction forces changed in relation to each other with age. Using this form of analysis, the data indicated that the older subjects were able to retain eccentric strength with age better than concentric strength in both muscle groups. In addition, in the hamstrings, eccentric strength was also retained better than isometric strength.

This tendency for eccentric strength to be retained better with age than other types of strength, despite still declining with age, has been shown previously in the quadriceps by Porter et al. (1995), who used the same analysis method, and Poulin et al. (1992) and Lindle et al. (1997), who normalised elderly values to the young mean. Vandervoort et al. (1990) reported a lower numerical decline in eccentric than concentric strength in the quadriceps and hamstrings, but these data were not subjected to statistical analysis.

Reasons for a reduced age effect upon eccentric contractions have been proposed in the literature. It has been suggested that an adaptive neural strategy to optimise eccentric activation may occur in older people, involving elimination of the eccentric inhibition normally seen in younger subjects (Enoka, 1996).

An *in vitro* study has also demonstrated that eccentric force declines more slowly with age in animals (Phillips et al. 1991) indicating that non-neural mechanisms may also play a part. Hortobagyi et al. (1995) has suggested that older subjects may have relatively greater eccentric strength due to increased passive resistance from greater concentrations of connective tissue or increased sarcomere strength (resulting from the previous elimination of weaker sarcomeres throughout life). Porter et al. (1997) and Hortobagyi et al. (1995) have also suggested that slower cross-bridge cycling may augment eccentric strength as the myosin heads will remain attached to actin over a longer duration, and thus the compliant S2 portion of the heads will be stretched further, increasing cross-bridge force (De Ruiter and De Haan 2001).

These data also show that in both hamstrings and the weaker quadriceps, isometric strength is retained better with age than concentric strength. In contrast, Harridge et al. (1995) did not note any age effect on isometric/concentric ratios in the plantarflexors, but these contractions were elicited by supramaximal electrical stimulation. One possible reason for better retention of isometric than concentric strength might be that any age-related slowing of the muscle will shift the force velocity curve to the left. This would mean that at a set concentric contraction speed older subjects will show more of an age-related concentric strength decline relative to young subjects than they would without the slowing effects. Since isometric contractions are not influenced by the force velocity curve, the isometric age decline would be less pronounced. Porter et al. (1997) have used a similar explanation to account for the greater loss of concentric compared to eccentric strength.

3.4.3 Differential age effects on eccentric strength at different speeds

It should be noted that throughout this thesis the relative terms fast (or faster) and slow (or slower) have been used to describe contractions at 50 and 150°sec^{-1}. However, 150°sec^{-1} is not truly fast in relation to the maximum speed of around 300°sec^{-1} seen in older subjects (Lanza et al. 2003). This is a limitation of this study that should be taken into account.

An important point highlighted by this study's results is that the speed of contraction may also influence the degree of eccentric strength loss with age. The averaged ratios of slow to fast eccentric strength show that the older retain faster speed eccentric strength better than slower speed eccentric strength. This was more pronounced in the hamstrings, as the numerical group differences in the individual quadriceps ratios were not significant.

This is the only study to report greater sparing of fast than slow eccentric strength with age in the hamstrings, as the hamstrings have not previously been studied in this context as far as is known. In the quadriceps, Poulin et al. (1992) also noted that eccentric strength only differed between ages at slower speeds. In contrast, Lindle et al. (1997) found that both slow and fast quadriceps eccentric contractions declined with age, and that there was no age effect on slow to fast eccentric strength ratios.

Theoretical considerations would tend to support the results seen in this study. The previously described effect of slower cross-bridge cycling in augmenting eccentric force may be further enhanced at higher eccentric velocities because the cross-bridges will be attached over an even longer movement distance, which further increases resistance to stretch in the S2 portion of the myosin cross-bridge head (De Ruiter and De Haan 2001, Porter et al. 1995). Hence it might be expected that older people, with a likely lower speed of cross-bridge cycling, might produce larger eccentric forces than expected at higher velocities, and thus retain eccentric strength better at these velocities.

The hamstring effect was not strongly significant (P=0.021) and in the light of the large number of variables considered, there is a moderate chance that this result was a type I error. The fact that this effect was only seen on the stronger leg supports this concern. However, averaging across muscles and legs did show a significant effect, and so these results offer some support for greater sparing of faster eccentric strength with age.

3.4.4 Differential age effects on concentric strength at different speeds

For concentric contractions, the finding that the young had a higher slow to fast concentric ratio in both legs of the hamstrings suggests that faster hamstring concentric contractions are more spared by age. This concurs with Aniansson et al. (1992) who reported that quadriceps concentric strength decreases with age were not seen at faster velocities. Delbaere and Bourgois (2003) noted a similar effect in women but not in men.

In contrast, Anianson et al. (1983) showed that women lost faster quadriceps concentric speed more rapidly. Previous findings in the dorsiflexors (Lanza et al. 2003) and the quadriceps (Lindle et al. 1997, Lanza et al. 2003) have reported no difference in the extent to which fast and slow concentric speed is lost with age. Some studies have reported significant age effects at all tested speeds in the quadriceps (Aniansson et al. 1986, Frontera et al. 2000a, Laforest et al. 1990) and plantarflexors (Cunningham et al. 1987) but differences between the declines at different speeds were not analysed statistically.

It has been suggested that faster concentric contractions should decline with age to a greater extent because of the concomitant slowing that occurs with age (Lindle et al. 1997). More specifically, the slower detachment of cross-bridges in the older subjects (Porter et al. 1995) should theoretically lead to lower forces at higher concentric velocities, as the S2 portions will be unloaded to a greater extent by the longer distance that the cross-bridges are attached. This effect will add to the age-related static weakness that is independent of velocity, so that faster contractions should decline more steeply with age than slower contractions. Hence the results in this study and others (Aniansson et al. 1992, Delbaere and Bourgois 2003) are paradoxical.

If these results are taken at face value, they have unexpected implications. Given that 150°.sec^{-1} is roughly half the maximal isokinetic velocity attained by older subjects (Lanza et al. 2003) and thus considerably less than half of Vmax, then the convergence of the young and older force velocity curves at the higher velocity of 150°.sec^{-1} implies a subsequent cross-over before Vmax is reached, with the older subjects having greater Vmax. Although there is controversy about the role of selective type II fibre loss with age (Aniansson et al. 1986, Lexell et al. 1988, Phillips et al. 1993c, Hortobagyi et al. 1995) and reductions in velocity in single fibres (Larrsson et al. 1997, Trappe et al. 2003) there is no evidence in the literature of an increase in velocity with age, and so this result can be regarded as extremely unlikely to have arisen physiologically.

Delbaere and Bourgois (2003) suggested that lower velocities may decline more with age as such movements are not functional and so are used less often. This is not a convincing argument as there is little evidence that the young engage in more non-functional slow activities. The possibility that inertial artefacts could explain results can be excluded as such artefacts were not observed. There is some evidence that neural inhibition may occur at the slowest isokinetic speeds, reducing torque (Osternig et al. 1986,Aargaard et al. 2000), and so if this occurred to a greater extent in the older subjects this might explain the wider gap in torque between ages at 50 rather than 150° sec^{-1} . However, there appears to be no evidence of such an age difference in inhibition in the literature. Training may reduce such inhibition (Aargaard et al. 2000), and so if the younger subjects were more trained this might also explain results. Although data on baseline training status were not collected, it is not unreasonable to suppose that more of the younger group were trained. It should be noted that although this study's finding was bilaterally consistent for the hamstrings, it was not highly significant ($P<0.01$) in either leg ($P=0.035$ weaker leg, $P=0.011$ stronger leg), and given the number of variables considered there is some risk of a type I error. Moreover when values were averaged across muscles and sides, the groups did not differ. Hence this study's results should be interpreted with caution.

3.4.5 Age effects on rectus femoris CSA

Ultrasound showed a pronounced decrease in rectus femoris muscle CSA with age in both legs. This concurs with other studies using computed tomography (CT) (Overend et al. 1992, Klitgaard et al. 1990a) and MRI (Jubrias et al. 1997, Klein et al. 2001) and supports the belief that part of the weakness associated with ageing is due to sarcopenia.

Ultrasound scanning has been shown to agree well with MRI scanning in terms of estimation of muscle CSA (Walton et al. 1997, Reeves et al. 2004a). However, in contrast to CT, MRI, or possibly other ultrasound models, the machine used in this study did not have sufficient image quality to permit accurate estimates of intramuscular non-contractile tissue area, which may increase in older muscle and therefore ideally should be excluded from the measurement (Phillips et al. 1991, Jubrias et al. 1997, Klein et al. 2001). Hence true muscle CSA in older subjects may be exaggerated when using ultrasound. However, this does not threaten the validity of conclusions, as such a bias will have worked against the observed effect of lower muscle CSA in the older subjects rather than assisting it.

ACSA rather than PCSA was measured in this study, as measurement of PCSA requires accurate measurement of muscle volume with MRI or CT (Narici et al. 2003), which were unavailable. The pennation angle in the RF has been reported as 22-25° in young people (Gianini et al. (1989), and therefore might be lower in older people. This could lead to the ACSA in the younger subjects being a greater underestimation of true PCSA than in the older subjects. However, for the same reasons given in the preceding paragraph, this would not affect the validity of conclusions. The rectus femoris was chosen as the muscle to be measured as it was small enough for the image to fit on the apparatus screen, and therefore facilitated area measurement. However, its smallness may have led to greater relative measurement error than might occur with a larger muscle such as vastus lateralis, and this drawback should be taken into account. Larger quadriceps muscles, or the whole quadriceps group, could have been measured by summating separate images spanning the entire muscle or muscle group, using a metal grid on the skin surface as a marker for the points of overlap, and future work, using similar measurement apparatus, should attempt this.

3.4.6 Age effects on index of F/CSA

The F/CSA values from this study cannot be compared to those in other studies as a true force value should measure the tension in the quadriceps muscle itself, accounting for co-activation of antagonists and levels of voluntary activation. Also, the CSA should be of the whole muscle group from which the muscle force was measured. Hence the F/CSA values given are merely an index of specific force. However, they are probably valid for group comparison purposes, as the ratio of the force exerted at the force transducer at the ankle to the quadriceps tensile force should be fairly constant across individuals and ages if it is assumed that the ratio of shank length to quadriceps moment arm (centre of knee rotation to patella distance) is also fairly constant across individuals and ages (Appendix 8). Voluntary activation and levels of co-activation were also similar between groups, further reducing the risk of confounding. Also, the ratio of RF CSA to overall quadriceps area is probably constant as is explained below.

Overall, there was an unexpected significant trend for the older subjects to have a higher index of F/CSA across the different specific force variables, suggesting that the decline in strength with age is due to sarcopenia and not reduced specific force. More intriguingly, it suggests that specific force may improve in older people as a compensation for sarcopenia. However, as far as is known, a tendency for older people to have higher F/CSA has not been observed previously, and previous studies have either shown increased quadriceps specific force in younger people (Young et al. 1985, Klitgaard et al. 1990a, Overend et al. 1992, Jubrias et al. 1997) or no difference (Young et al. 1984, Overend et al. 1992, Hakkinen and Hakkinen 1991, Kent-Braun and Ng 1999, Frontera et al. 2000b). There were no differences between young and old in activation or co-activation that could explain this result, and so it could relate to greater intrinsic strength (i.e. at the level of the contractile apparatus) in the older subjects. However, there is no evidence in the literature that intrinsic strength increases with age.

Methodological factors do not seem to explain these results. The use of rectus femoris CSA as an index of overall quadriceps area could explain these results if there were a smaller ratio of rectus femoris to whole quadriceps CSA in older people. However, Trappe et al. (2001) showed that the ratios do not change with age. Another factor could be the use of ultrasound to measure CSA, and the use of ACSA. As explained, both may lead to a relative over-estimation of CSA and thus under-estimation of specific strength in older people. However, because of the direction of this potential bias – a lessening of specific strength in the older subjects - this possible methodological drawback does not threaten the conclusion that older subjects had greater specific force. Importantly, however, the lack of significant differences between groups for individual isometric and concentric variables may be due to this methodology.

Another possibility was that the young group were less motivated. Although the voluntary activation test results were similar between groups, and thus appear to show that the young were equally motivated, it is possible that motivation will have been stepped-up in such a test where effort was being overtly measured. Differences in motivation are conceivable between two groups where variations in attitude and background may have varied more between, than within, groups. Lower motivation in the young would explain their lower F/CSA, since force would be lower than expected whilst CSA would be unaffected.

When variables were considered individually, only eccentric specific force demonstrated a significantly greater F/CSA in older subjects. This is the first measure of eccentric F/CSA in the quadriceps as far as is known. This specific result may be explained by eccentric force being partially influenced by non-muscular factors such as greater non-contractile resistance to stretch (Hortobagyi et al. 1995), or lower levels of neural inhibition (Enoka 1996). Since eccentric strength may depend less on muscle CSA than the other contaction types, there may be some justification for not including the eccentric specific force variables in the Binomial analysis. However, exclusion of the eccentric variables did not alter the overall result, though the significance decreased from <0.01 to <0.05.

The fact that isometric strength and rectus femoris CSA do not correlate except for the young men suggests that the variability in women's and the older men's strength was influenced more by intrinsic and neurological than morphological factors. Morse et al. (2004) reached the same conclusion in response to their analogous finding of a strong correlation between Triceps surae volume and plantarflexion strength in young but not old men. This implies a greater variation in specific force in women and older people than young men. Reasons for this are yet to be ascertained.

3.4.7 Contribution of co-activation of antagonists to weakness

The results suggest that co-activation of antagonists during isometric and isokinetic contractions of the quadriceps and hamstrings do not differ between young and old, and thus do not contribute to age differences in measured strength. The isometric results contrast with all previous isometric findings during knee extension (Izqierdo et al. 1999, Macaluso et al. 2002, Tracy and Enoka 2002) but concur with previous findings during knee flexion (Macaluso et al. 2002). Isokinetic results contrast with previous findings during concentric knee extension (Izquierdo et al. 1999) but agree with previous findings during eccentric and concentric knee extension (Tracy and Enoka 2002).

There are no obvious reasons for the conflict between this study's results and most of the results in the literature. The subject numbers and age groups were similar to those in the other studies. A potential source of error was cross-talk between agonist and antagonist electrodes. Cross-talk would tend to give the impression of more co-activation, and if the young were more affected by cross talk than the older subjects, this could explain the conflicting results. Cross talk might occur to a greater extent in subjects with greater amounts of subcutaneous fat, but body mass data does not suggest that the young were more obese. Hence an age effect on cross talk effects is unlikely. Moreover, Klein et al. (2001) did check for cross-talk in similar methodological studies and found no evidence that this was occurring.

Another reason for the different isometric results may relate to the method of calculating co-activation. In the isometric contractions, co-activation was expressed as the IEMG of the antagonist (during the agonist's maximum contraction) normalised to the IEMG of that same antagonist when it was acting as agonist *at the same angle as the optimum angle of the previous agonist.* This was not performed in the other studies; instead the antagonist EMG was simply normalised to the IEMG of the same antagonist when it was acting at an agonist at its *own* optimum angle. Such a calculation of co-activation using IEMG data derived from two different angles might skew results, as IEMG varies with the joint angle (Babault et al. 2003), and may possibly thus explain the differing findings.

3.4.8 Contribution of voluntary activation to weakness

Despite the relatively large group sizes used in this study, the percutaneous twitch interpolation technique was unable to discern any differences in quadriceps motor unit recruitment between age groups. This implies that strength decreases in the quadriceps with age are unlikely to be due to lower motor unit recruitment.

Hurley et al (1998) and Roos et al. (1999) also used a comparable single or double twitch interpolation technique and gained similar results in the quadriceps. In contrast, Stackhouse et al. (2001) and Stevens et al. (2001), who showed higher muscle activation for young subjects, used the high frequency stimuli approach. The differences in findings are therefore likely to be due to the differing techniques (Stevens et al. 2003).

The technique of percutaneous twitch superimposition (Belanger and McComas 1981, Rutherford et al. 1986) normally involves the supramaximal stimulation of the muscle with discrete pulses or doublets to evoke a muscle twitch. The amplitude of this twitch at rest represents the summation of twitch force from stimulated motor units. During a maximal voluntary contraction, the amplitude of the twitch above the voluntary force will only be produced by those stimulated motor units that have not been voluntarily activated. Thus the ratio of the twitch during a maximum contraction to the resting twitch will represent the proportion of motor units that are not voluntarily activated. Although this technique may only stimulate around 60% of the quadriceps muscle, it has been shown to give similar results to supramaximal stimulation (at 1Hz) of the femoral nerve (Rutherford et al. 1986). In contrast, the high frequency stimuli approach will lead to a fully fused tetanic muscle response. Thus the amplitude of any electrically evoked response during an MVC will represent a deficit due to both incomplete motor unit recruitment and/or suboptimal frequency coding (Kent Braun and Le Blanc 1996). Hence the activation measured by high frequency stimulation is distinct from that measured by the percutaneous twitch interpolation technique, the latter only measuring the deficit of motor unit recruitment.

In conclusion then, it can be stated that the level of motor unit recruitment alone does not appear to be responsible for reduced muscle force in older people, based on the results from this study. However, results from the studies using the 100Hz train of stimuli approach (Stackhouse et al. 2001, Stevens et al. 2001) suggest that suboptimal rate coding may contribute to the weakness seen in older people. In concert with this, decreased discharge rates of motor units have been noted in older men (Connelly et al. 1999) and such decreased rates have also been cited as a reason for reduced single fibre force (Kamen et al. 1995, Jubrias et al. 1997). Jubrias et al. (1997) stated, however, that lower frequencies will only lead to lower forces if the potentiating longer relaxation times in older muscles do not fully compensate for this.

Some limitations of the method used should be noted. A pre- rather than post-MVC resting twitch was used to establish the level of maximum activation deficit. On reflection, this may have been in error, as the twitch evoked during the MVC will possibly have been potentiated by the muscle activity. Comparing this to the non-potentiated pre-MVC twitch may tend to lead to lower estimations of MVC activation deficit than comparison with a potentiated post-MVC twitch, such as used by Morse and colleagues (2004). In addition, the use of twitch doublets may be more sensitive in detecting age differences (Morse et al. 2004) than the single twitches used in this study, possibly because of a higher signal to noise ratio (Behm et al. 2001). It should also be noted that this study used maximally-tolerated rather than strictly supramaximal stimuli. Whilst a supramaximal stimulus will recruit all motor axons that are within the spatial range of the stimulus, a maximally tolerated stimulus may fail to recruit the smaller axons with the highest stimulus threshold (Trimble and Enoka 1991). Hence calculation of activation using non-supramaximal stimuli will only measure the proportion of the larger axons that are voluntarily recruited. This technique may therefore underestimate activation levels, as the smaller axoned units are likely to have been voluntarily activated, by Henneman's size principle (Henneman 1981). However, this limitation may not affect the validity of results, as the older subjects may have had a greater proportion of slower units (with smaller motor axons) and may therefore have endured a greater underestimation of activation than the young.

3.4.9 Power

The finding that younger subjects had greater unilateral lower limb peak extensor power in both legs during a single movement concurs with the one other finding in the literature (Skelton et al. 1994). Other studies have also observed a decrease with age when power was measured in both legs simultaneously (Bosco and Komi 1980, Izquierdo et al. 1999, Runge et al. 2004, Petrella et al. 2005). These results may be comparable to unilateral results, as although bilateral muscle performance is often less than the sum of both limbs working unilaterally (Taniguchi 1997, Kurugati et al. 2005), age does not seem to affect the degree to which this phenomenon occurs (Kurugati et al. 2005). Hence it can be assumed that results from bilateral studies are indices of unilateral age differences and do not merely reflect the effects of ageing on the bilateral to unilateral power relationship.

One possible drawback of this study was the failure to measure power with an inertial load that would approximate that required for maximal power production. The Nottingham Power Rig has a flywheel of fixed inertial load. For maximal power testing to be truly valid, the variation of power with contraction force needs to be considered (Pearson et al. 2004, Pearson et al. 2001, Macaluso and De Vito 2003). Power is maximal at a certain percentage of the unique maximal force of an individual, so if all subjects are using a power-measuring device that has a fixed inertial load, most will be unable to achieve true peak power as this fixed load is unlikely to equal most subjects' optimal load for maximal power production. If two groups are being studied with very different strength levels, one group may be favoured in terms of power production. For example, young subjects will have greater strength and so will have an optimal power production at a higher level of resistance.

In contrast, several studies have measured peak power at various inertial force levels, thus ensuring that a force level leading to an approximation of true peak power was used, and peak power was found to occur at about 50-60% of maximum isometric force for young and old (Macaluso and De Vito 2003, Izquierdo et al. 1999, Petrella et al. 2005). In all these studies, comparison of true peak power still showed the young to be more powerful.

Regardless, the Nottingham power rig has been designed to test older subjects and its flywheel inertia is optimised for weaker subjects. This means that the younger subjects and not the older would be at a disadvantage. Hence this does not threaten the conclusions that the young were more powerful, as had optimisation of the load been performed, the young would probably have performed even better.

The finding of reduced power in the older subjects is to be expected, given the reductions in strength. What is of more interest is any difference in decline between strength and power – a greater decline in power would imply a concomitant decrease in contraction velocity as well. The similar proportions of strength and power in the older and younger subjects suggest that power may be lost at similar rate to strength. This conflicts with previous studies showing that declines in power are greater (Skelton et al. 1994). This implies that either contraction velocity does not change with age or that the lack of difference was due to a biased comparison of strength and power, leg extensor power involving the hamstrings and gluteus maximus as well as the quadriceps. The latter is more likely given previous findings of decreased fast muscle mass (Klitgaard et al. 1989, Lexell and Downham 1992, Trappe et al. 2003).

However, since concentric strength is also an index of power, comparison of the the concentric to isometric ratios in young and old provided an alternative meaure of the relative changes of power and strength with age. These showed that power is indeed lost at a greater rate than strength, indicating a loss of speed with age.

3.4.10 Asymmetry of strength and power

This is the first study to directly explore asymmetry of strength, muscle area and power in young and older subjects. The results demonstrate no group differences in asymmetry of muscle area and lower limb power. The only group differences in strength asymmetry were observed during the fast concentric contraction of the quadriceps, the slow concentric contraction of the hamstrings and averaged concentric values.

The lack of significant differences between ages for asymmetry of isometric and eccentric strength, power and rectus femoris CSA suggests two possibilities. First, if the losses in strength, power or muscle size occur sporadically on either side of the body then they must individually be small losses in order that broad symmetry is maintained. This is in contrast to the losses seen in motor neurone diseases such as ALS (Mochizuki et al. 1995) where large losses occur focally, leading to significant asymmetry of strength and power. These results are also consistent with a more general loss of strength and power that evolves bilaterally.

Present theories of age-related losses would tend to support both ideas. Loss of anterior horn cells may be random and sporadic, which would support the former possibility.

However, reductions in muscle fibre diameter, or reductions in specific strength may well be more universal and so support the second possibility. In conclusion, the two mechanisms probably occur together, with no effect on symmetry.

The difference between groups in concentric asymmetry is surprising in the light of the lack of differences in isometric asymmetry and power asymmetry. If there is a greater loss of faster fibres with age, then any asymmetry in this process might partially explain the co-existence of these and the isometric results through an asymmetry of speed affecting concentric rather than isometric asymmetry. However, the faster concentric hamstrings and lower limb power asymmetry did not differ between groups, which suggest no differences in asymmetry of speed between groups. Further studies are required to confirm these results.

3.5 Conclusions

This study supports previous limited evidence of a slower decline of eccentric strength and greater relative sparing of faster than slower eccentric strength. It has also suggested there are no age differences in antagonist co-activation, and that asymmetry of concentric strength may increase with age. There is also some suggestion that central activation may not differ with age, but the methodological drawbacks should be bourne in mind.

There were some unexpected results that could not be explained by methodology. For example, fast concentric strength declined more slowly with age than slow concentric strength in the hamstrings, which implied a possible increase in contraction velocity with age. Similarly, there was a significant trend for F/CSA to increase with age. These findings oppose current knowledge of ageing and further work is required to confirm them.

4 Muscle strength, muscle power and asymmetry of strength and power: the association with falls risk in community dwelling elderly

4.1 Introduction

This section will examine evidence that medically unexplained falls in community-dwelling older people are associated with deficits in strength, power and symmetry in the knee and ankle flexors and extensors. These factors are potentially important as they may be remediable (Chapter 9).

Unexplained falls have been associated with other age-related motor deficits such as reduced joint flexibility and poorer trunk equilibrium reactions (Allum et al. 2002) but these are outside the scope of this study and will not be considered.

4.1.1 Strength

A number of retrospective studies have shown that falls in community-dwelling elderly people appear to be associated with weakness in the leg extensors (De Rekeneire et al. 2003, Gehlsen and Whaley 1990), knee extensors (Lord et al. 1992, Lord et al. 1999, MacRae et al. 1992), knee flexors (Robinson et al. 2004, MacRae et al. 1992), ankle dorsiflexors (Robinson et al. 2004, Roma et al. 2001, Daubney and Culham 1999, Skelton et al. 2002, Studenski et al. 1991, MacRae et al. 1992), ankle plantarflexors (Robinson et al. 2004), hip flexors (Robinson et al. 2004) hip abductors (MacRae et al. 1992) and hip extensors (Daubney and Culham 1999). However, several studies have not detected strength differences between fallers and non-fallers in the quadriceps (Schwendner et al. 1997, Daubney and Culham, 1999, Skelton et al. 2002, Melzer et al. 2004, Robinson et al. 2004), the hamstrings and plantarflexors (Skelton et al. 2002, Daubney and Culham 1999, Melzer et al. 2004) or the dorsiflexors (Melzer et al. 2004). These discrepancies may be related to the differing types, speeds and angles of contraction used during testing. A study utilising both isometric and isokinetic tests at a variety of angles and speeds is therefore required.

Other retrospective studies have also been carried out on institutionalised people. Lord et al. (1991) found no difference in dorsiflexion or quadriceps strength between fallers and non-fallers. In contrast, falling has been associated with weakness in the knee extensors (Whipple et al. 1987, Ikezoe et al. 2003) and flexors, ankle dorsiflexors and plantarflexors and trunk flexors (Whipple et al. 1987). However, this sample may have different health characteristics to the "healthy" fallers under focus in this study.

There is some literature documenting a reduction in falling associated with strength training (Mulrow et al. 1994, Campbell et al. 1997, Campbell et al 1999, Robertson et al. 2001). However such training may also improve power and possibly other potential falls-related factors such as steadiness (Laidlaw et al. 1999, Hortobagyi et al. 2001).

In retrospective studies it is difficult to distinguish between the causes and effects of falls, and furthermore reports of the number of falls may be unreliable (Graafmans et al. 1996). Prospective studies for community dwelling people of >70 years old have been performed. Some have demonstrated an association between subsequent falls and lower strength in the quadriceps (Takazawa et al. 2003, Campbell et al. 1990, Luukinen et al. 1995, Tinetti et al. 1995), dorsiflexiors (Takazawa et al. 2003) and plantarflexors (Luukinen et al. 1995) but no studies used isokinetic measures and often used subjective manual muscle testing (Tinetti et al. 1995, Luukinen et al. 1995) or functional measures of strength (Campbell et al. 1990). A meta-analysis of prospective studies has shown that lower strength is associated with falling (Moreland et al. 2004), but this included falls in institutionalised elderly and many of the included studies did not measure strength objectively.

4.1.2 Power output

Only two retrospective studies on community-dwelling subjects have investigated power output in fallers. Skelton et al. (2002) found that fallers were less powerful (when normalised for body mass) than non-fallers. It has also been noted that the power output in the uninjured limb of elderly people with fall-related hip fractures was 70% less than that of a group of age matched non-fallers (Levy et al. 1994, cited by Skelton et al. 2002). However, it is possible that subjects who have had recent surgery would not be able to produce pre-fracture power levels in the non-fractured limb due to pain and reduced stabilisation from the operated side.

Whipple et al. (1987) and Fleming et al. (1991) found that nursing home residents who fell had lower power than non-fallers, but the latter study did not use age-matched non-fallers. Moreover, evidence from such a population may not relate to the older community-dwelling population as studied here.

Few prospective studies have been carried out. Kemoün et al. (2002) found that elderly people who fell in the subsequent year had lower eccentric hip extension power during the stance phase of gait, estimated from kinematic analysis. This does not, however, indicate that fallers have reduced maximal power. Graafmans et al. (1996) and Tinetti et al. (1995) found that elderly people with lower power measured by timed chair stands were also more likely to fall in the subsequent period.

Some biomechanical studies involving simulated trips, slips or forward falls in young subjects have suggested that a certain level of lower limb power is required for recovery (Eng et al. 1994, Schillings et al. 2000, Pijnappels et al. 2005a, Tang et al. 1998). These suggestions have been based on kinematic, force plate and EMG data. However, these studies do not directly support a link between power and falling in the elderly, as the elderly may use different recovery strategies. Related studies have shown that older subjects use similar strategies, but produce less estimated power and recover with more difficulty (Tang and Woolacott 1998, Thelen et al. 2000), strengthening the association between power and falls. Hall et al. (1999) reported contradictory results, with no difference in magnitude or rate of development of ankle torque between young and old, but subjects were stationary when exposed to anteroposterior floor translations and so this may lack relevance to activities of daily living.

Lower proportions of Type II fibres have also been associated with falls risk. Elderly patients undergoing hip fracture repair had significant reductions in Type II fibre areas compared with age-matched controls (Aniansson et al. 1984). The actual numbers of fibres of each type were unchanged. The surgical subjects were described as having 'fresh' hip fractures, which suggests that the difference in Type II fibre area preceded, rather than resulted from, the injury. Contraction speed is a determinant of power and smaller fast fibres will mean slower contraction and thus lower power (Whipple et al. 1987).

4.1.3 Asymmetry

Only one study has investigated the link between falls and lower limb asymmetry of strength and power in the elderly: Skelton et al. (2002) found that leg extension power asymmetry in elderly community-dwelling fallers was greater than in age-matched non-fallers. It was also found that strength asymmetry did not differ between the groups.

4.1.4 Implications

Retrospective studies of the association between reduced strength, power, symmetry and falls are incomplete, limited or conflicting. There is also a need for further prospective studies, but these were beyond the resources of this project. Hence the hypotheses of this study were that, compared with age-matched healthy controls, elderly fallers have:

1. Reduced lower limb strength

2. Reduced lower limb power

3. Increased asymmetry of strength

4. Increased asymmetry of power

4.2 Methods

4.2.1 Subjects

The older non-faller and faller groups only (chapter 2, page 21) participated in this part of the study. In this chapter the older non-fallers will be referred to as "non-fallers".

4.2.2 Muscle tests

4.2.2.i Isometric strength

Knee extensor and flexor isometric strength was measured as described in Chapter 3. In addition, ankle plantarflexion and dorsiflexion isometric strength was measured on the same equipment. The subjects lay in supine, with the knees fully extended and supported by a pillow (Fig. 4.1). The axis of ankle flexion was aligned by eye with the lever arm axis of rotation and the foot was placed in the purpose-built ankle plantarflexion/dorsiflexion measurement apparatus. After several practice attempts, peak isometric forces at 0 (plantargrade), 10, 20 and 30° of plantarflexion were obtained for both muscle groups.

Each contraction lasted for 3 seconds, with a minimum rest period of 5 seconds between each. Two repetitions at each angle were allowed for each muscle group.

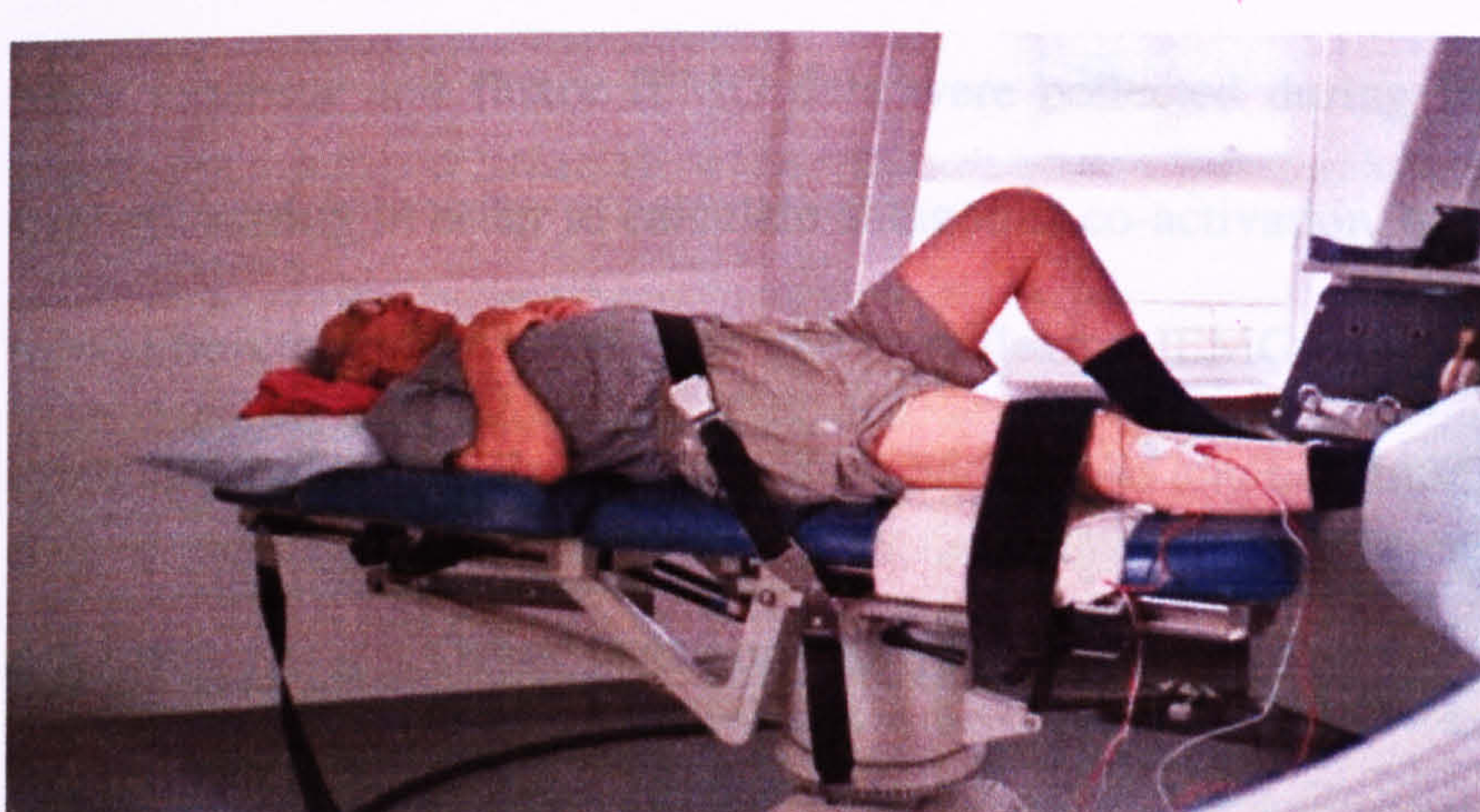


Fig. 4.1 Positioning of subjects for isometric dorsiflexion and plantarflexion strength testing.

4.2.2.ii Isokinetic strength

Knee extensor and flexor isokinetic strength was measured as described in chapter 3. In addition, ankle plantarflexion and dorsiflexion isokinetic strength was measured on the same equipment. Isokinetic and isometric ankle measurements were made in the same positions. Measurements were made at two angular velocities: 50 and $150°.sec^{-1}$ in random order. The testing range was 0 - 20° plantarflexion. A minimum rest period of 5 seconds was allowed between each trial. For each muscle, a minimum of 3 trials were performed until values decreased. The first trials were regarded as practice attempts. Test-retest reliability of isokinetic ankle testing (as measured by Pearson correlation) has been measured at R=0.92 at $60°.sec^{-1}$ and R=0.87 at $120°.sec^{-1}$ for plantarflexion and at R=0.90 at $60°.sec^{-1}$ and R=0.75 at $120°.sec^{-1}$ for dorsiflexion on the Cybex II (Morris-Chatta et al. 1994). These values were obtained from older subjects with an inter-test interval of 6 months, so ageing effects may have affected results.

Dorsiflexion and plantarflexion forces were not corrected for gravity, as the gravity effects on ankle muscle forces in supine were considered negligible.

4.2.2.iii Co-activation

Knee extensor and flexor IEMG data were collected during isometric and isokinetic strength testing in order to calculate antagonist co-activation, as described in Chapter 3. In addition, ankle plantarflexion and dorsiflexion IEMG data were collected using the same electrodes over the gastrocnemius and tibialis anterior muscle bellies.

4.2.2.iv Other tests and analyses

Rectus femoris CSA, lower limb power and quadriceps voluntary activation were measured as described in Chapter 3. Ratios of the strength of different contraction types (slow speed) and ratios of slow to fast concentric and eccentric force were calculated.

4.2.3 Activity level measurements

Activity monitors (Actigraph Inc., USA) were used to establish if groups were equivalent in terms of activity levels. These matchbox-sized monitors contain an accelerometer that records vertical whole body movements.

Subjects were asked to wear the monitors strapped to their right hip during waking hours for 7 consecutive days and to continue with normal activities. They were asked to record times when the device was removed, or when on wheeled transport (which could cause spurious vertical movements of the device in the absence of physical activity). This procedure was only conducted on a sample of the elderly fallers (n=12) and non-fallers (n=24), as the devices were not available during the period when the majority of young subjects were being tested.

Analysis was performed by a collaborator at Bristol University (Mark Davis, Dept. Exercise and Health Science) for whom the activity data were also being collected. Moderate activity or above was defined as that producing >1952 counts minute^{-1} (Morse et al. 2004), and the number of minutes per day at that level was recorded as the activity variable. Corrections were made for any periods when on wheeled transport.

4.2.4 Statistical analysis

Inter-group comparison of the knee extensor and flexor data was performed as described in Chapter 2. Inter-group analyses concerning dorsiflexor and plantarflexor data were analysed with a GLM univariate ANOVA only across the faller and non faller groups and so no post hoc tests were used. A GLM univariate ANOVA was used instead of independent t tests to permit correction for sex differences. An additional analysis for frequent fallers (3 or more falls) was carried out in the same way, with those falling less than 3 times being eliminated from the analysis rather than joining the non-faller group.

A Pearson product correlation was carried out to analyse the relationship between number of falls in female fallers and strength, power, RF CSA, co-activation and asymmetry.

4.3 Results

4.3.1 Baseline group characteristics and potential confounding variables.

The elderly fallers and non-fallers did not differ in activity levels, body mass, or age-corrected height squared. An independent t test showed that age did not differ between the groups (P>0.05) (Table 4.1).

	Fallers		Non-fallers		Post hoc p (FvNF)	Adjust-ments
	n	Mean (SE)	n	Mean (SE)		
Body mass (kg)	34	70.76 (1.97)	44	70.44 (1.64)	>0.05	sex
Height (m)	34	1.64 (0.01)	44	1.68 (.001)	.049	sex
Age corrected height (m)	34	1.69 (0.01)	44	1.73 (.001)	>0.05	sex
Age corrected height^2 (m^2)	34	2.88 (.043)	44	2.96 (.035)	>0.05	sex
Male	5		15			
Female	30		29			
Age (yrs)	35	76.4 (.78)	44	75.9 (.61)	>0.05	
Activity level $	12	28.0 (4.3)	24	27.6 (3.2)	>0.05*	sex

Table 4.1 Baseline characteristics of elderly fallers and non-fallers. The groups did not differ in activity levels, age, weight or corrected height squared. (F=Fallers, NF=Non-fallers) $ = minutes per day at >moderate activity level (>1952 counts/min on actigraph monitor). *As activity levels were only recorded for fallers and non-fallers, only these groups were included in the univariate analysis.

4.3.2 Interactions with sex

Sex interacted (P<0.05) with CSA, power and strength/power variables, and all strength variables except the following: isometric plantarflexion at 20 and 30° bilaterally; all isokinetic plantarflexion variables; concentric dorsiflexion at 150°.sec^{-1} bilaterally; and averaged concentric dorsiflexion and concentric and eccentric plantarflexion values. Male sex increased the value. For the only other variables interacting with sex, being male led to an increase in the ratio of slow to fast hamstring concentric strength in the strong leg, slow to fast dorsiflexion concentric and eccentric strength bilaterally, plantarflexion isometric to eccentric on the strong leg and a decrease in the ratio of isometric to eccentric hamstring strength in the weak leg (P<0.05). Group comparisons were corrected for these interactions.

4.3.3 Strength

4.3.3.i Isometric

There was a clear pattern for fallers to be weaker across the isometric variables; the mean of 46/48 variables were numerically lower in the fallers. A Binomial test showed this was highly significant (P<0.01). For individual variables, the fallers had significantly lower isometric strength than non-fallers in the quadriceps and dorsiflexors at a limited number of angles, and in the hamstrings and plantarflexors at a fuller range. In all muscle groups apart from the hamstrings, peak forces occurred at the same angle for both groups, indicating no difference in the length-tension relationship. However in the hamstrings the fallers appeared to show peak forces at a shorter muscle length, particularly in the weaker leg (Figs. 4.2-4.9).

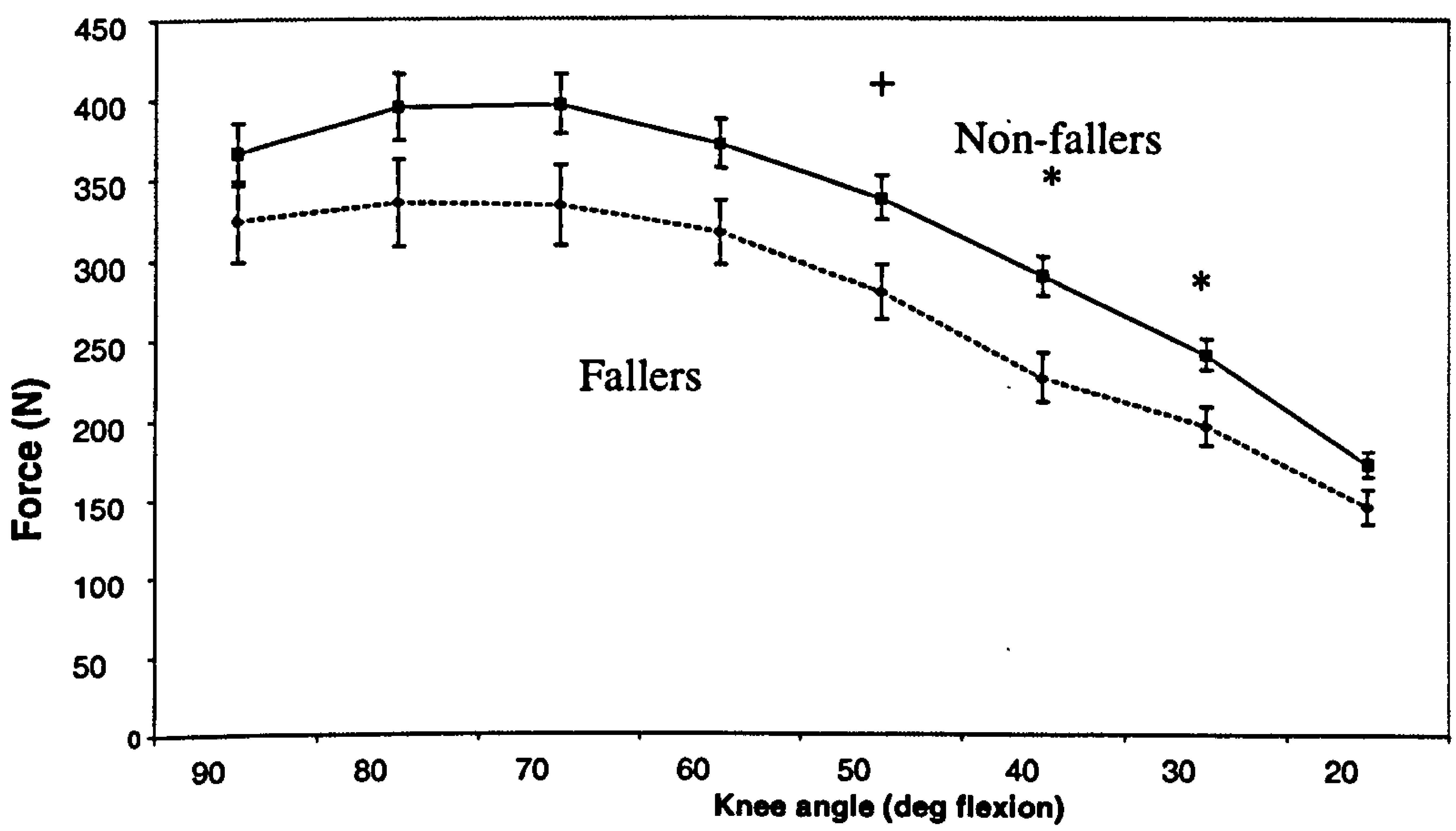


Fig. 4.2 Isometric quadriceps strength in the stronger leg in fallers (n=23-28) and non-fallers (n=39-40). * P<0.01 . + P<0.05. (N = Newton)

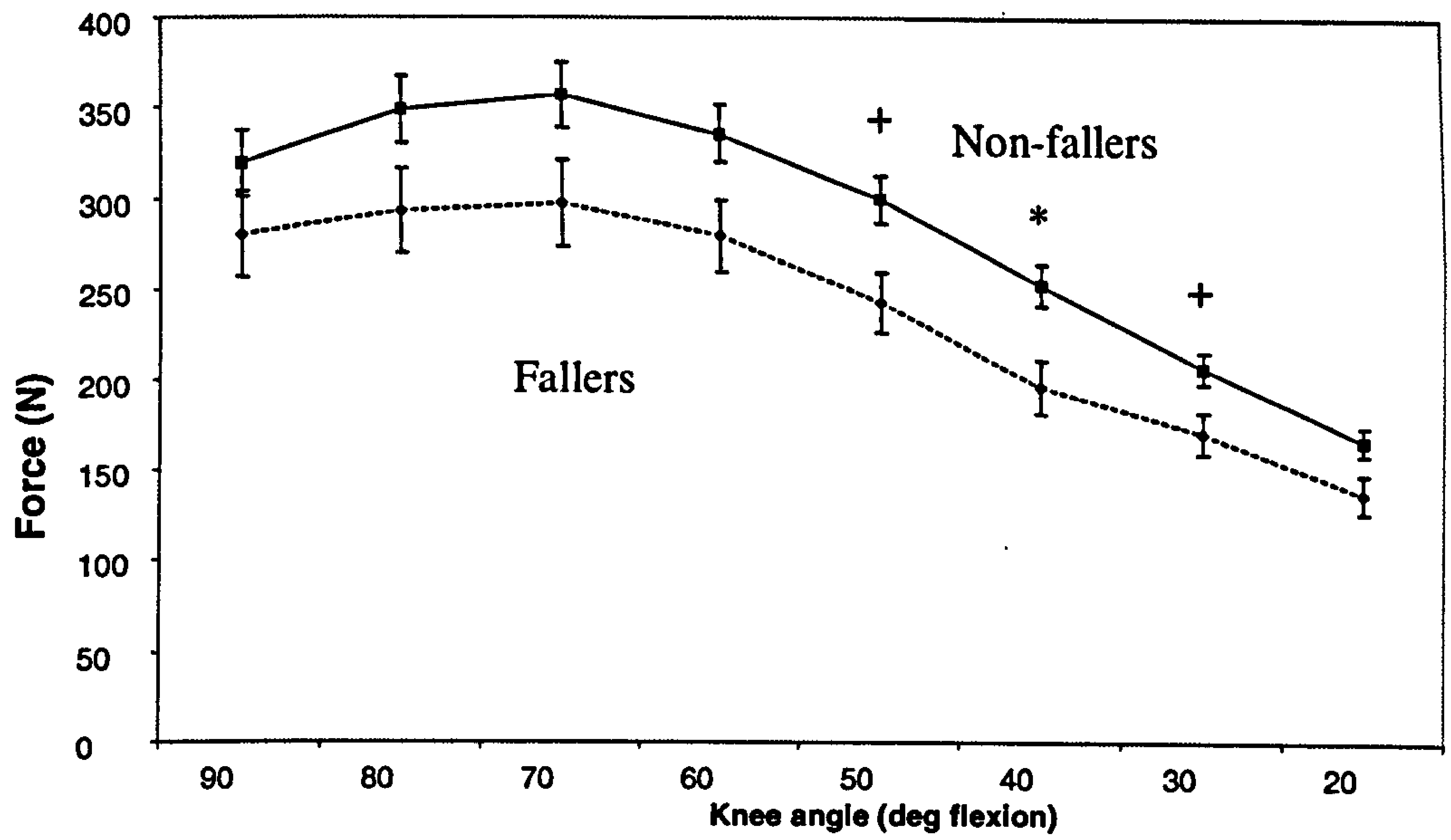


Fig. 4.3 Isometric quadriceps strength in the weaker leg in fallers (n=23-28) and non-fallers (n=39-40). * P<0.01 + P<0.05. (N = Newton)

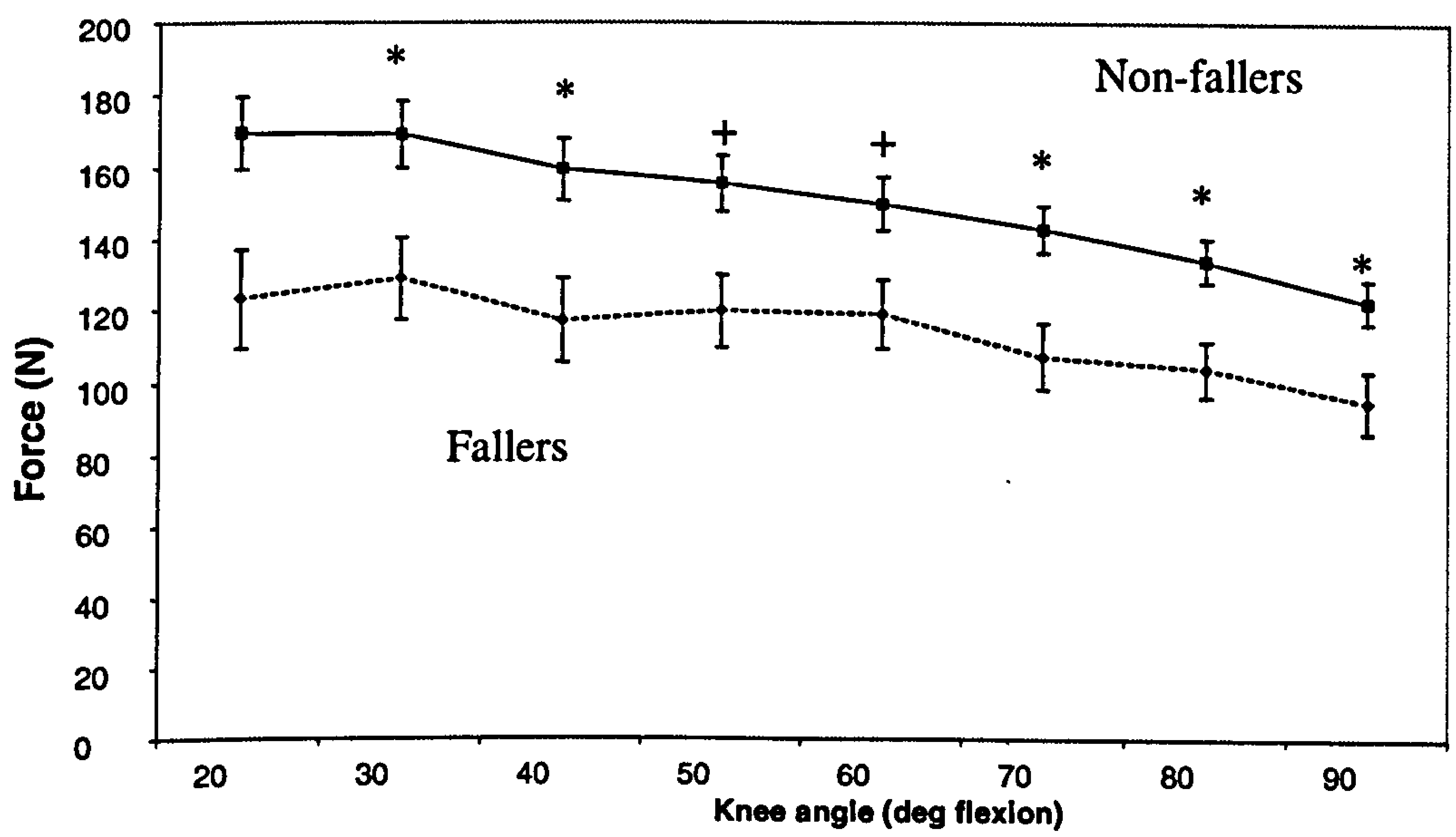


Fig. 4.4 Isometric hamstring strength in the stronger leg in elderly fallers (n=23-27) and non-fallers(n=38-40). * P<0.01 +P<0.05. (N = Newton)

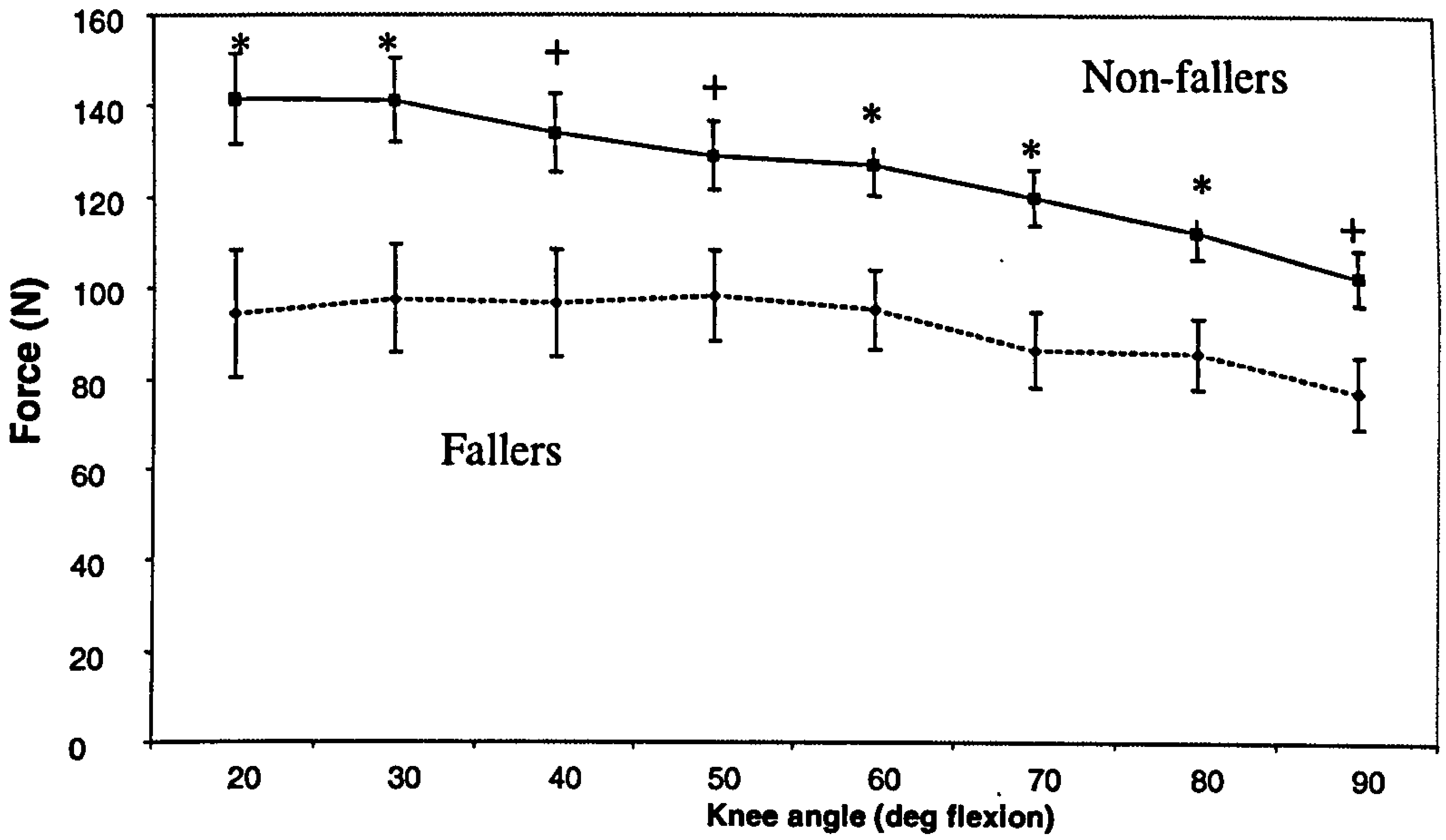


Fig. 4.5 Isometric hamstring strength in the weaker leg in fallers (n=23-27) and non-fallers (n=38-40) * P<0.01 + P<0.05. (N = Newton)

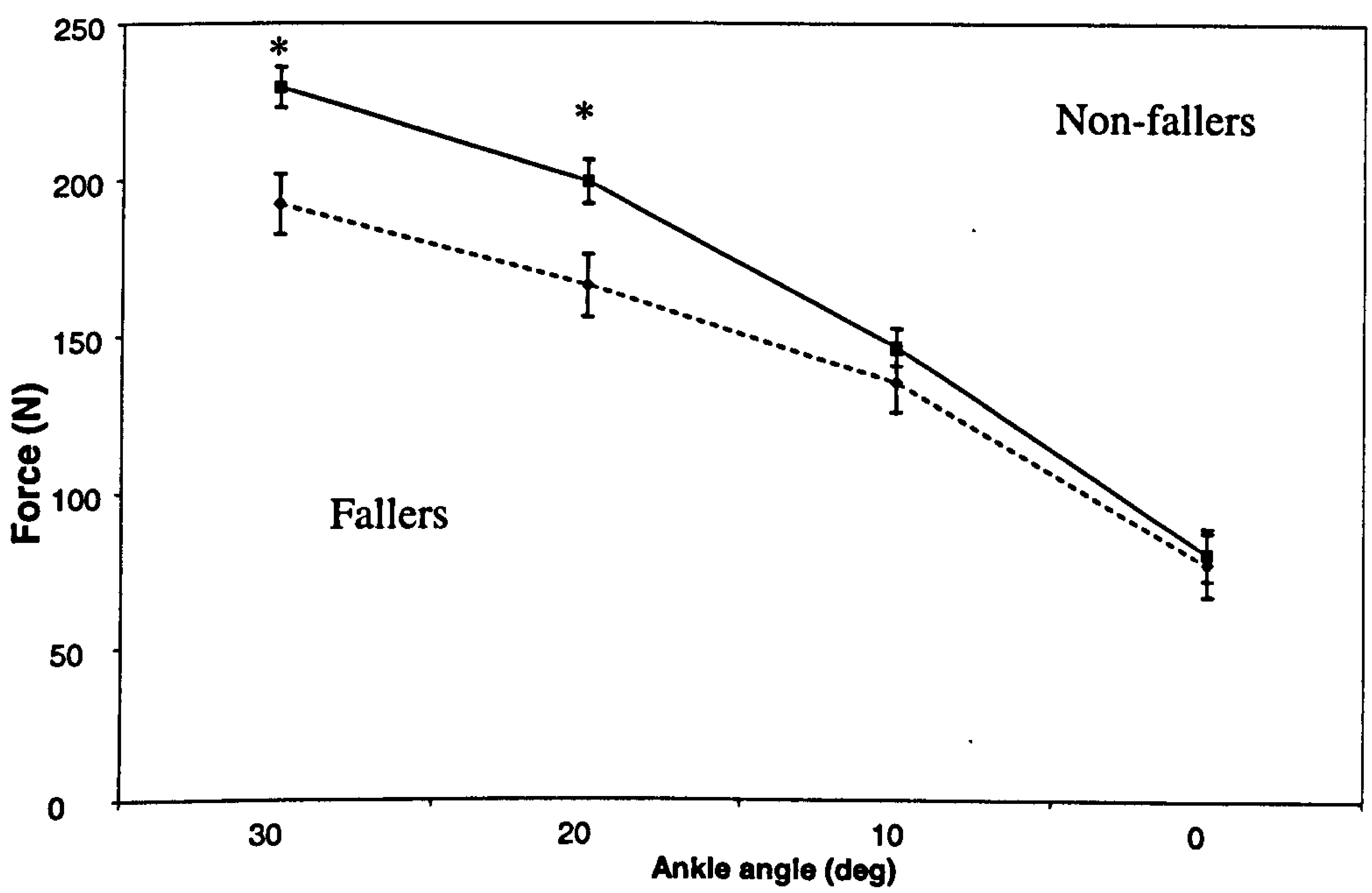


Fig. 4.6 Isometric dorsiflexion strength in the stronger leg in fallers (n=11-25) and non-fallers (n=11-41). *P<0.01 (N = Newton)

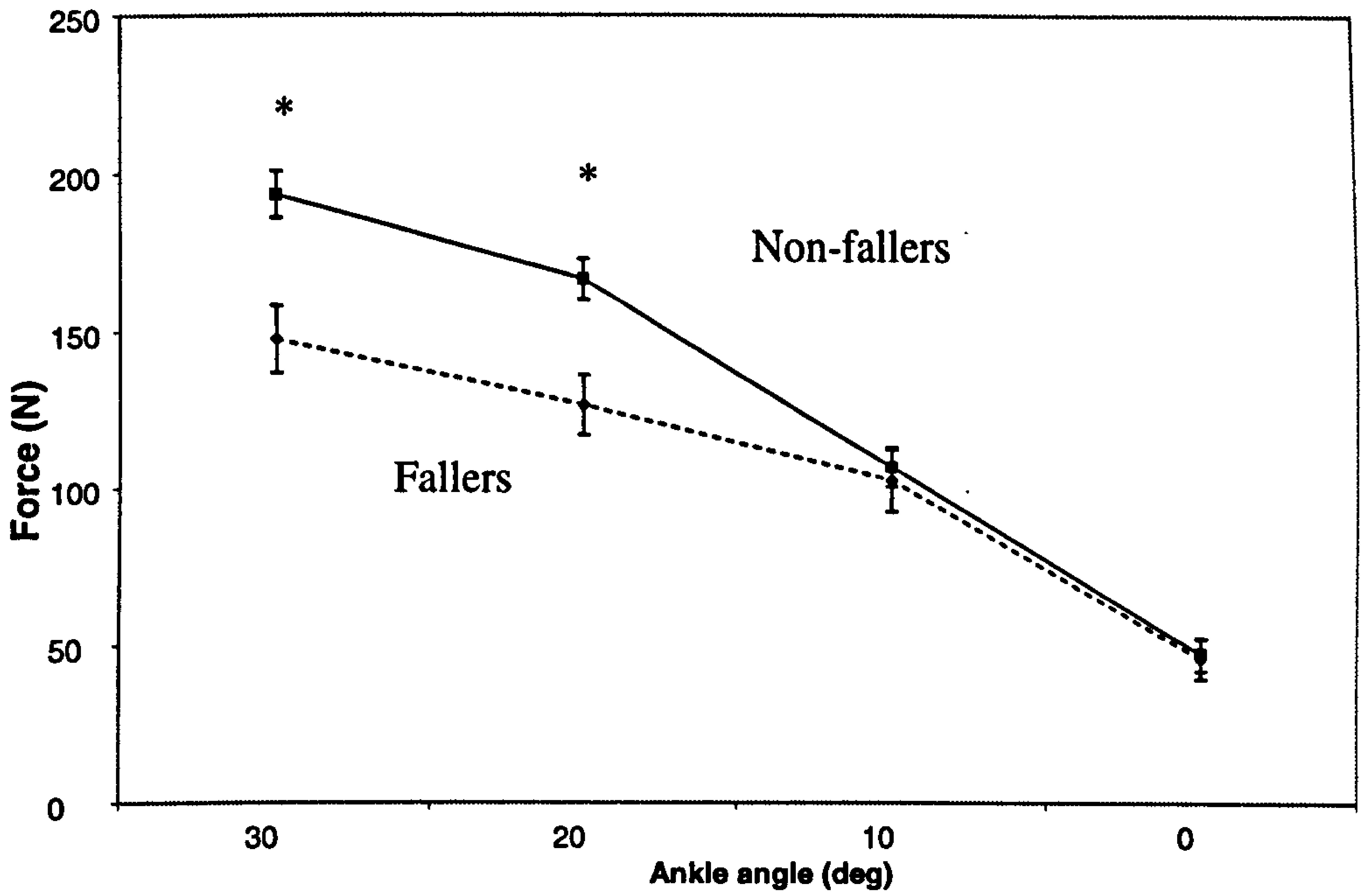


Fig. 4.7 Isometric dorsiflexion strength in the weaker leg in fallers (n=11-25) and non-fallers (n=11-41) * P<0.01 (N = Newton)

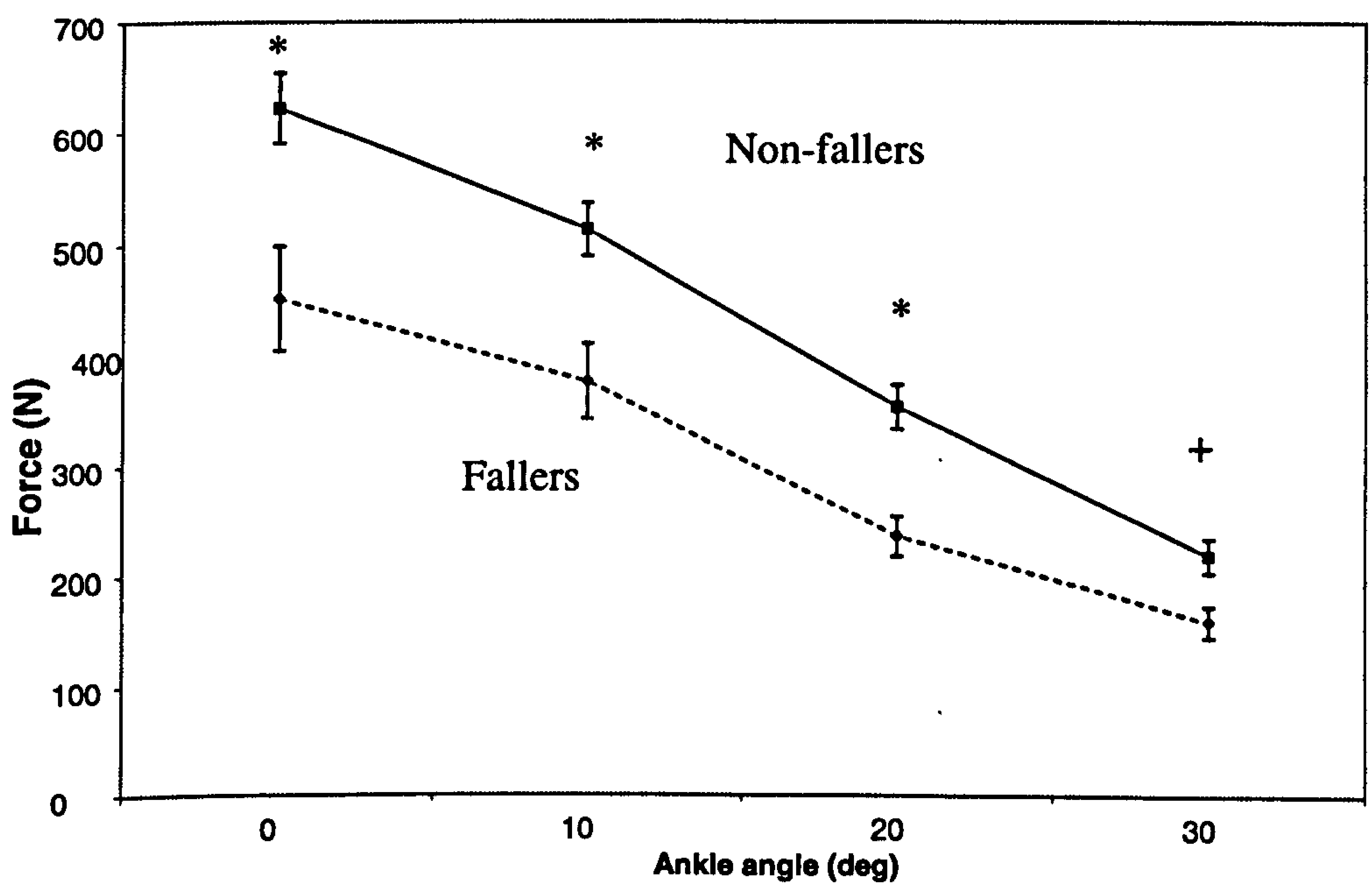


Fig. 4.8 Isometric plantarflexion strength in the stronger leg in fallers (n=21-25) and non-fallers (n=39-41) * P<0.01 + P<0.05. (N = Newton)

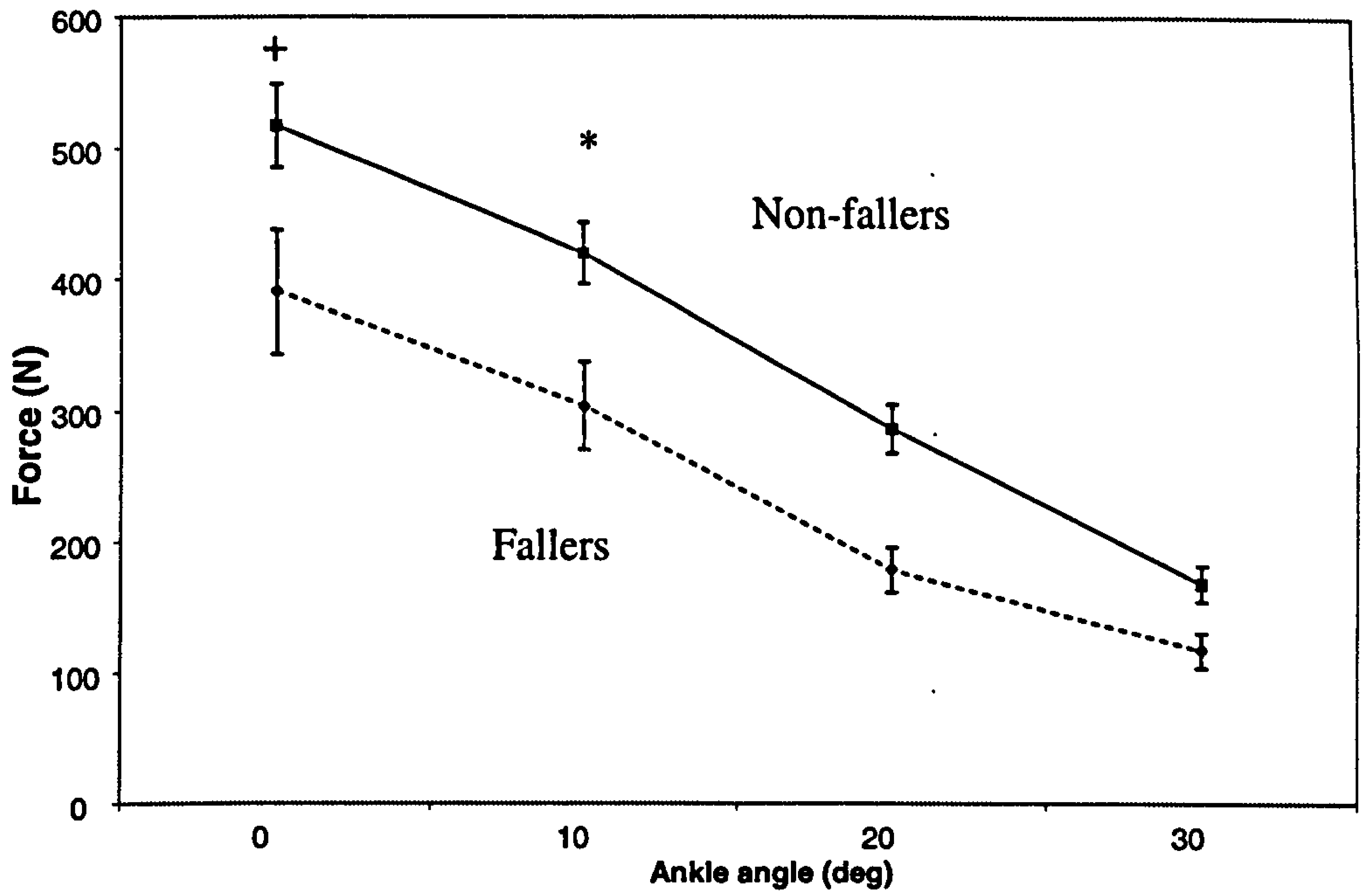


Fig. 4.9 Isometric plantarflexion strength in the weaker leg in fallers (n=21-25) and non-fallers. (n=39-41) * $P<0.01$ + $P<0.05$. (N = Newton)

4.3.3.ii Isokinetic strength

The Binomial test showed there was a significant trend for isokinetic strength to be lower in fallers across all concentric ($P<0.01$) and all eccentric ($P<0.01$) variables. When variables were analysed individually, fallers had significantly weaker hamstring eccentric strength in the weaker leg at $50^{\circ}.sec^{-1}$, weaker dorsiflexor eccentric strength in the stronger leg at $50^{\circ}.sec^{-1}$, and all concentric and eccentric plantarflexor contractions except the eccentric contraction at $150^{\circ}.sec^{-1}$ on the stronger leg. The groups did not differ in other isokinetic variables. When variables for each muscle group were averaged across legs and speeds, only the plantarflexors showed lower concentric and eccentric strength in the fallers (Table 4.2).

		Quadriceps		Hamstrings		Dorsiflex.		Plantarflex.	
		Fallers	Non fallers	Fallers	Non fallers	Fallers	Non fallers	Fallers	Non fallers
C	50 strong	257 (25)	310 (22)	105 (12)	134 (11)	78 (7)	89 (5)	166 (16)+	231 (18)
	50 weak	204 (25)	257 (22)	77 (12)	100 (10)	63 (6)	72 (4)	124 (14)*	192 (16)
	150 strong	220 (18)	243 (15)	119 (10)	142 (8)	65 (2)	64 (3)	125 (10)+	158 (10)
	150 weak	187 (18)	196 (15)	89 (10)	112 (9)	56 (2)	55 (2)	94 (8)*	127 (9)
	Av	227 (28)	254 (17)	99 (10)	123 (9)	61 (4)	68 (3)	127(15)*	179 (11)
E	50 strong	429 (32)	480 (26)	171 (14)	205 (12)	217 (17)+	269 (12)	437 (29)*	578 (28)
	50 weak	357 (28)	410 (24)	127 (13)+	170 (11)	196 (17)	232 (11)	342 (34)*	486 (27)
	150 strong	416 (27)	472 (22)	190 (11)	210 (9)	194 (16)	216 (10)	386 (27)	455 (22)
	150 weak	358 (27)	404 (22)	159 (10)	179 (8)	175 (16)	184 (10)	298 (27)+	377 (24)
	Av	397 (26)	448 (21)	163 (10)	191 (9)	200 (17)	218 (10)	358 (28)*	477 (20)
	n	24-25	31-32	24-25	32-34	14-19	31-35	21-22	34-35

Table 4.2. Isokinetic strength (N) (mean(SE)) in fallers and non-fallers. + = significant difference between groups ($P<0.05$) *= significant difference between groups ($P<0.01$) Av=average, C=concentric, E= eccentric, 50=50°.sec^{-1}, 150=150°.sec^{-1}

4.3.3.iii Force ratios between different contraction types and speeds

Averaging of values across legs and muscles showed that non-fallers had a greater ratio of slow to fast concentric (fallers 1.08±0.03, non-fallers 1.28± 0.03, $P<0.001$) and slow to fast eccentric strength (fallers 0.99±0.03, non-fallers 1.15±0.04, $P<0.001$). Other averaged ratios did not differ.

When variables were analysed individually, non-fallers had a significantly higher ratio of slow to fast dorsiflexion eccentric contraction strength in the strong leg, slow to fast hamstring eccentric strength on the weak leg, and slow to fast quadriceps and plantarflexion concentric force on the weak leg. Other related variables did not vary between groups (Table 4.3).

	Strong leg		Weak leg	
	Fallers	Non-fallers	Fallers	Non-fallers
Q I/C	1.39 (0.09)	1.31 (0.07)	1.55 (0.09)	1.44 (0.09)
Q I/E	0.8 (0.04)	0.81 (0.03)	0.82 (0.04)	0.88 (0.07)
Q C/E	0.6 (0.02)	0.65 (0.03)	0.57 (0.03)	0.63 (0.03)
Q slow/fast C	1.23 (0.04)	1.33 (0.05)	1.13 (0.04)*	1.41 (0.06)
Q slow/fast E	1.02 (0.03)	1.02 (0.02)	0.99 (0.04)	1.03 (0.03)
H I/C	1.44 (0.12)	1.41 (0.08)	1.61 (0.14)	1.71 (0.12)
H I/E	0.75 (0.04)	0.82 (0.03)	0.74 (0.05)	0.8 (0.04)
H C/E	0.58 (0.04)	0.62 (0.03)	0.56 (0.06)	0.55 (0.04)
H slow/fast C	0.92 (0.05)	0.83 (0.05)	0.76 (0.07)	0.86 (0.07)
H slow/fast E	0.83 (0.05)	0.94 (0.03)	0.74 (0.04)*	0.91 (0.03)
DF I/C	2.58 (0.25)	2.67 (0.15)	2.39 (0.19)	2.74 (0.17)
DF I/E	0.92 (0.06)	0.82 (0.02)	0.79 (0.06)	0.83 (0.03)
DF C/E	0.40 (0.02)	0.35 (0.02)	0.37 (0.02)	0.34 (0.02)
DF slow/fast C	1.25 (0.09)	1.41 (0.06)	1.18 (0.09)	1.34 (0.06)
DF slow/fast E	1.12 (0.06)+	1.26 (0.04)	1.15 (0.06)	1.26 (0.04)
PF I/C	2.49 (0.16)	2.78 (0.21)	2.78 (0.19)	2.77 (0.22)
PF I/E	0.98 (0.06)	1.04 (0.04)	1.00 (0.05)	0.99 (0.06)
PF C/E	0.39 (0.03)	0.41 (0.02)	0.38 (0.02)	0.40 (0.02)
PF slow/fast C	1.34 (0.10)	1.44 (0.05)	1.28 (0.07)+	1.46 (0.05)
PF slow/fast E	1.17 (0.06)	1.31 (0.05)	1.19 (0.07)	1.44 (0.16)

Table 4.3. Force ratios (mean (SE)) between different contraction types and speeds in both groups. * = fallers significantly different to non-fallers ($p<0.01$), + = fallers significantly different to non-fallers ($p<0.05$), Q=quadriceps, H=hamstrings, S=dorsiflexion, PF=plantarflexion, C=concentric, E=eccentric, I=isometric. Fallers n=11-24, Non-fallers n=31-33.

4.3.3.iv Strength normalised to body mass

The results were very similar to the non-normalised results in terms of significant differences between groups. There were only 5 departures from the non-normalised results: quadriceps strength at 60° bilaterally and hamstring strength at 20° on the stronger leg were greater for non-fallers in this analysis, whilst isokinetic plantarflexion did not show any group differences at 150°.sec^{-1} concentrically and eccentrically in the stronger leg. Overall, correcting for body mass did not qualitatively change the strength results, and strength normalised to body mass will not be considered further. Means and SE data are in Appendix 2.

4.3.3.v Normalised strength – corrected height squared

The results were similar to the non-normalised results in terms of significant differences between groups. All hamstring and dorsiflexor results were identical. In contrast to the non-normalised analysis, isometric quadriceps strength was greater in non-fallers at 20° on the weak leg, but not at 30 and 50° bilaterally, and plantarflexion strength at 30° bilaterally was similar in both groups. Concentric plantarflexion was similar at $50°.sec^{-1}$ in the strong leg and $150°.sec^{-1}$ bilaterally, and eccentric plantarflexion was similar at $50°.sec^{-1}$ in the strong leg and $150°.sec^{-1}$ in the weak leg. However, eccentric strength at $150°.sec^{-1}$ was higher for non-fallers in this analysis. Again, correcting for corrected height squared did not qualitatively change most of the strength results. Strength normalised to corrected height squared will therefore not be considered further. Means and SE data are in Appendix 2.

4.3.4 Rectus femoris cross-sectional area

There was no difference in rectus femoris CSA between groups in either leg (Table 4.4).

	Fallers	Non-fallers
RF CSA larger leg	4.7 (0.5)	4.6 (0.3)
RF CSA smaller leg	4.0 (0.4)	3.8 (0.3)

Table 4.4 Rectus Femoris CSA in fallers (n=17-29) and non-fallers (n=15-29). There were no differences between groups.

A correlation analysis between the CSA (in strong and weak legs) and quadriceps isometric strength at the strongest angle of 80° (in strong and weak legs) was performed. This correlation was performed separately for each combination of sex and group to prevent group or sex effects influencing results. The group of male fallers was not considered as the number of subjects in this group was small (max 4).

In both legs for all 3 subgroups (male non-fallers, female fallers, female non-fallers), there were no significant correlations (Table 4.5).

	Stronger leg	Weaker leg
Male non-fallers	R=0.61, P=0.11, n=8	R=0.67, P=0.10, n=8
Female non-fallers	R=0.11, P=0.65, n=19	R=0.21, P=0.40, n=19
Female fallers	R=0.11, P=0.75, n=11	R=-0.19, P=0.59, n=11

Table 4.5. Correlation between CSA and isometric strength in both legs. There were no significant correlations.

4.3.5 Index of F/CSA

A Binomial test showed there was a significant trend across all 10 F/CSA variables ($P<0.01$) and across all 6 isometric and concentric F/CSA variables ($P<0.05$) for the non-fallers to have greater F/CSA. However, when analysed individually, F/CSA variables did not differ between groups (Fig. 4.10).

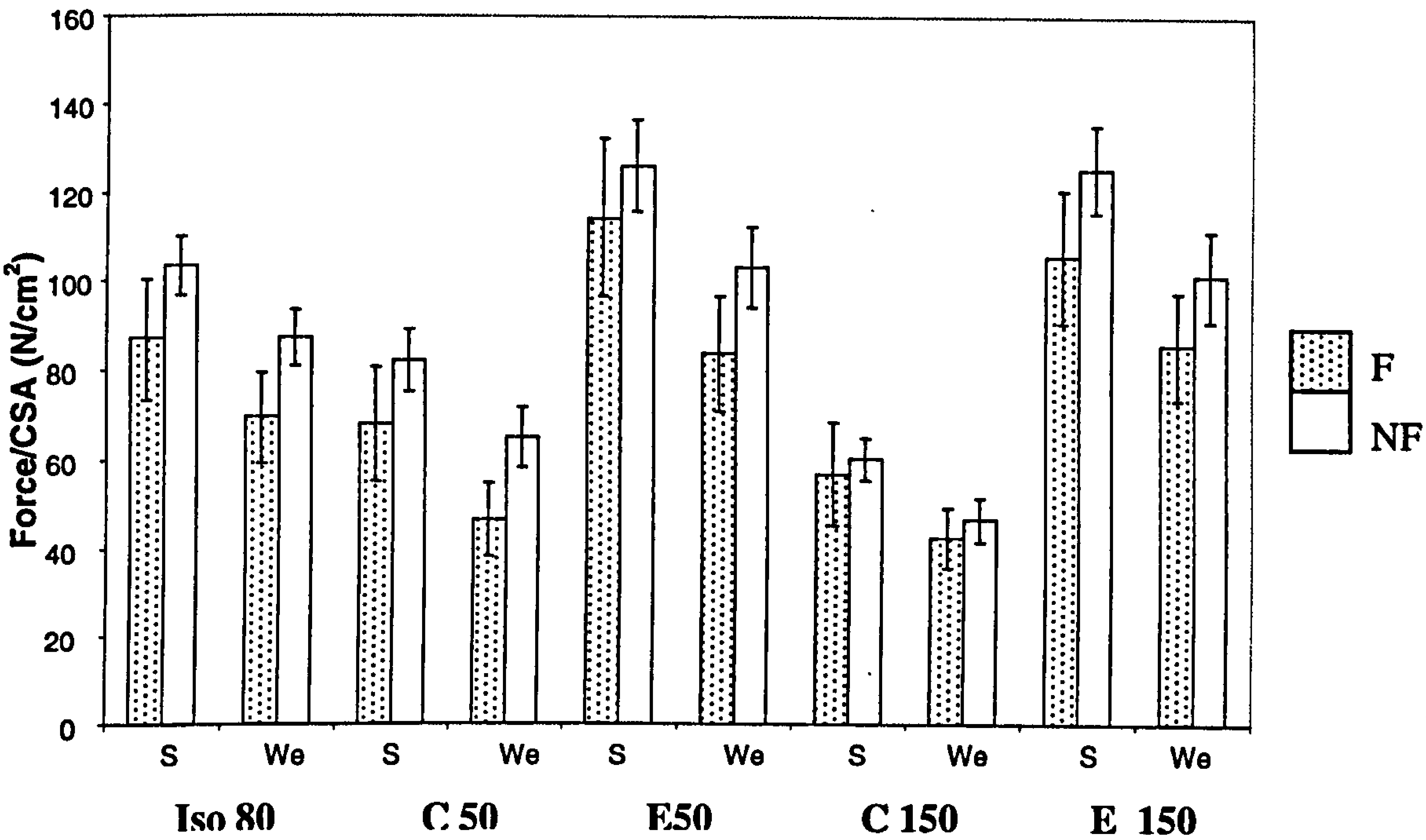


Fig. 4.10 Quadriceps isometric, concentric and eccentric F/CSA in fallers (F, n=10-11) and non-fallers (NF, n=20-25). The groups did not differ in F/CSA. S=strong, We=weak, C=concentric, E=eccentric, Iso=isometric, numbers refer to speed for isokinetic contractions and angle for isometric contractions.

4.3.6 Voluntary muscle activation

There were no differences in voluntary activation levels between elderly fallers and elderly non-fallers in either the weaker or stronger legs (Table 4.6).

	Fallers	Non-fallers
% activation weaker leg	92.8 (3.1)	97.9 (1.0)
% activation stronger leg	92.6 (4.1)	95.6 (2.0)

Table 4.6. Voluntary activation in fallers (n=22-23) and non-fallers (n=31).There were no differences between groups.

4.3.7 Co-activation levels

Overall, there was a definite trend for the fallers to have greater isometric co-activation, with all isometric variables across legs and muscles having greater numerical means in the fallers. A Binomial test showed this was significant (P<0.01). However there were no such trends for co-activation variables during concentric or eccentric contractions. When variables were analysed individually, the fallers had higher co-activation of their plantarflexors during eccentric dorsiflexion in the weaker leg than non-fallers. There were no other individual differences between groups (Table 4.7).

		Strong leg		Weak leg	
Contraction type of agonist	Antagonist	Fallers	Non-fallers	Fallers	Non-fallers
Isometric	Quads	0.181 (0.06)	0.122 (0.03)	0.221 (0.09)	0.175 (0.05)
	Hams	0.227 (0.03)	0.136 (0.02)	0.416 (0.09)	0.241 (0.05)
	Dorsi	0.209 (0.04)	0.172 (0.02)	0.242 (0.06)	0.227 (0.03)
	Plantar	0.18 (0.04)	0.162 (0.03)	0.244 (0.05)	0.186 (0.03)
Concentric	Quads	0.187 (0.07)	0.215 (0.07)	0.112 (0.02)	0.125 (0.02)
	Hams	0.169 (0.04)	0.224 (0.04)	0.276 (0.07)	0.206 (0.04)
	Dorsi	0.339 (0.08)	0.246 (0.03)	0.301 (0.09)	0.248 (0.03)
	Plantar	0.25 (0.05)	0.186 (0.04)	0.257 (0.06)	0.2 (0.04)
Eccentric	Quads	0.144 (0.05)	0.151 (0.04)	0.11 (0.03)	0.125 (0.03)
	Hams	0.233 (0.03)	0.168 (0.04)	0.221 (0.05)	0.276 (0.06)
	Dorsi	0.244 (0.07)	0.215 (0.02)	0.273 (0.12)	0.262 (0.03)
	Plantar	0.223 (0.06)	0.196 (0.05)	0.288 (0.04)+	0.167 (0.02)

Table 4.7 Relative antagonist activation (full activation = 1) during maximal agonist contraction in fallers (n=8-15) and non-fallers (n=18-25). + = P<0.05.

4.3.8 Power

Fallers and non-fallers did not differ in absolute power for either the most or least powerful legs, or power normalised to body mass for the least powerful leg. However, the non-fallers had significantly greater normalised power in the more powerful leg (Table 4.8).

		Fallers	Non-fallers
Absolute Power (W)	Most powerful leg	150.4 (12.6)	179.7 (9.8)
	Least powerful leg	131.9 (12.5)	157.3 (9.8)
Normalised Power (W/kg)	Most powerful leg	2.03 (0.15)	2.48 (0.17)+
	Least powerful leg	1.77 (0.15)	2.16 (0.12)

Table 4.8 Absolute and normalised power [mean(SE)] in fallers (n=29) and non-fallers (n=43). + = P<0.05.

When isometric quadriceps strength was divided by power there were no differences between groups. Sex interacted with these variables.

Strength/Power	Fallers	Non-fallers
Stronger leg	2.7 (0.16)	2.6 (0.13)
Weaker leg	2.6 (0.15)	2.4 (0.12)

Table 4.9 Isometric quadriceps strength(N) / power(W) in fallers (n=27) and non-fallers (n=38) . There were no differences between groups.

4.3.9 Asymmetry of strength

There was a clear trend for fallers to have greater isometric asymmetry than non-fallers across 21/24 isometric variables. The Binomial test showed this was highly significant (P<0.01). No such trends were noted for the concentric or eccentric variables.

When variables were analysed individually, fallers had greater isometric strength asymmetry than non-fallers in the hamstrings at 30 and 60° knee flexion, and the dorsiflexors at 20 and 30° plantarflexion only. Fallers also had greater isokinetic asymmetry than non-fallers in the eccentric hamstring contraction at $50^{\circ}.sec^{-1}$ but there were no other isokinetic asymmetry differences. There were no differences in rectus femoris cross-sectional area asymmetry or lower limb extensor power asymmetry (Tables 4.10-4.11).

	Angle (deg) or speed (°.sec^{-1})	Quadriceps		Hamstrings		Dorsiflexion		Plantarflexion	
		Fallers	Non-fallers	Fallers	Non-fallers	Fallers	Non-fallers	Fallers	Non-fallers
Iso	90	14.9 (2.1)	12.5 (1.2)	20.7 (3.0)	17.5 (2.6)	-	-	-	-
	80	12.8 (2.0)	11.4 (1.3)	19.6 (3.8)	17.5 (2.1)	-	-	-	-
	70	13.4 (2.0)	10.7 (1.2)	20.5 (3.4)	16.8 (1.8)	-	-	-	-
	60	12.5 (2.1)	9.3 (1.2)	23.4 (4.1)+	15.1 (1.7)	-	-	-	-
	50	12.3 (2.1)	10.3 (1.6)	19.0 (2.8)	17.2 (2.3)	-	-	-	-
	40	14.4 (3.0)	12.1 (1.3)	17.2 (3.0)	16.5 (2.1)	-	-	-	-
	30	12.0 (2.2)	12.7 (1.6)	22.6 (3.1)+	14.8 (2.1)	22.6 (3.3) +	15.3 (2.0)	26.6 (4.3)	22.6 (2.6)
	20	14.3 (2.9)	11.8 (1.7)	22.5 (3.1)	16.4 (2.0)	26.8 (4.0) +	17.3 (2.0)	25.4 (3.9)	18.1 (2.1)
	10	-	-	-	-	25.4 (4.7)	28.7 (3.0)	24.4 (3.8)	19.3 (2.4)
	0	-	-	-	-	39.2 (5.7)	34.7 (6.6)	19.1 (3.3)	19.9 (2.7)
C	50	27.6 (3.7)	21.1 (2.9)	28.3 (5.0)	27.5 (3.8)	16.2 (2.5)	19.0 (2.7)	23.8 (3.8)	17.0 (2.2)
	150	14.8 (1.8)	20.5 (2.6)	28.1 (3.8)	21.7 (2.9)	12.8 (1.9)	13.9 (2.1)	22.2 (3.5)	18.2 (2.4)
E	50	15.6 (2.5)	14.0 (2.4)	25.6 (2.6)*	15.0 (1.9)	9.6 (1.6)	15.4 (2.3)	24.1 (3.5)	16.4 (2.5)
	150	11.9 (1.6)	13.8 (2.4)	14.7 (2.0)	13.1 (1.5)	10.5 (2.8)	16.3 (2.2)	23.4 (2.9)	17.7 (3.0)

Table 4.10 Asymmetry [mean % diff between sides (SE)] during peak isometric, concentric and eccentric contractions in fallers (n=12-26) and non-fallers (n=30-40). *=fallers significantly different to non-fallers ($P<0.01$). + = fallers significantly different to non-fallers ($P<0.05$). iso = Isometric, C = concentric, E= eccentric.

	Fallers	Non-fallers
RF CSA	16.9 (3.3)	16.6 (1.9)
Power	18.1 (3.1)	13.8 (1.5)

Table 4.11. Asymmetry [mean % diff between sides (SE)] for RF CSA and lower limb extensor power in fallers (n=15-29) and non-fallers (n=29-43). There were no differences between groups.

4.3.10 Correlation between number of falls and muscle strength and power

There were no correlations between the number of falls in the women fallers and power or CSA (Table 4.12). Isometric force at all angles in all muscles, and isokinetic force in the quadriceps, dorsiflexors and plantarflexors, did not correlate with the number of falls. However, hamstring eccentric force at 50°.sec^{-1} on the weak leg, and at 150°.sec^{-1} on both legs, correlated positively with the number of falls. The level of co-activation in the weak concentric dorsiflexors during a maximal plantarflexion contraction correlated positively with the number of falls. Eccentric asymmetry at 150°.sec^{-1} in the dorsiflexors also correlated positively with the number of falls but in the plantarflexors the association was negative (table 4.13). A separate correlation for male fallers was not carried out as the number of subjects was too small.

		R	P	n
POWER	More powerful leg	-0.01	0.99	25
	Less powerful leg	<0.01	0.99	25
RF CSA	Larger leg	-0.15	0.61	13
	Smaller leg	-0.04	0.90	13

Table 4.12 Correlation of the number of falls in women with power and RF CSA. There were no significant correlations.

		Quads		Hams		Dorsiflex.		Plantarflex.	
	leg	n	R	n	R	n	R	n	R
Peak isometric	S	24	-0.19	24	0.1	23	0.31	23	-0.01
	We	24	-0.16	24	0.08	23	-0.07	23	0.02
Concentric $50°.sec^{-1}$	S	22	0.16	22	0.08	17	-0.16	20	-0.09
	We	22	-0.08	22	0.17	17	-0.17	20	<0.01
Eccentric $50°.sec^{-1}$	S	22	0.26	22	0.33	10	-0.25	19	0.25
	We	22	0.32	22	0.45+	10	-0.25	19	0.34
Concentric $150°.sec^{-1}$	S	22	0.24	21	0.31	17	0.09	20	-0.11
	We	22	0.23	21	0.27	17	0.18	20	0.06
Eccentric $150°.sec^{-1}$	S	22	0.28	21	0.47+	13	-0.11	19	0.01
	We	22	0.4	21	0.42+	13	-0.22	19	0.24
Isometric co-activation	S	14	-0.25	13	-0.24	12	-0.12	10	-0.47
	We	11	0.1	11	0.47	11	0.3	9	0.24
Concentric co-activation	S	11	0.29	10	-0.12	9	-0.2	9	-0.22
	We	11	-0.04	10	-0.24	9	0.87*	6	-0.39
Eccentric co-activation	S	12	0.19	11	0.06	7	0.16	7	-0.4
	We	10	-0.07	10	0.22	6	0.63	5	0.78
Asymmetry	Iso	23	-0.08	23	0.07	23	0.13	23	-0.26
	C 50	21	0.36	21	-0.1	17	0.04	20	-0.17
	C 150	21	-0.02	20	-0.07	17	-0.19	20	-0.11
	E 50	21	-0.22	21	-0.32	10	0.13	19	-0.32
	E 150	21	-0.12	20	-0.08	13	0.60+	19	-0.47+

Table 4.13 Correlation of the number of falls in female fallers with strength, antagonist co-activation and asymmetry. Only peak isometric angles are shown as no isometric variables were significant * $P<0.01$. + $P<0.05$ C=concentric, E=eccentric,50=$50°.sec^{-1}$,150=$150°.sec^{-1}$,S=stronger leg, We=weaker leg.

4.3.11 Results when only frequent fallers (≥3 falls) were considered.

When only those subjects having 3 or more falls were redefined as fallers, the non-fallers (which did not include those falling 1 or 2 times) had significantly higher isometric strength at 20° and 30° on the weaker dorsiflexors, and at all plantarflexor contractions except those on the weaker leg at 30°, 10° and 0°. Non-fallers had significantly higher isokinetic strength in all plantarflexion contractions except the eccentric contraction at $150°.sec^{-1}$ on the weaker leg. There were no differences in power, RF CSA, or quadriceps and hamstring isometric and isokinetic strength. Asymmetry was greater in the fallers in power and in isometric hamstring strength at 60°, and dorsiflexion strength at 30° and 20°. Co-activation was also greater in the fallers in the weaker hamstrings during maximal quadriceps isometric contractions, in the weaker dorsiflexors during maximal concentric plantarflexion, and in the weaker plantarflexors during maximal eccentric dorsiflexion. This additional analysis was performed in exactly the same way as the previous analyses. Sex did not interact with asymmetry, co-activation, power, CSA or activation ($P>0.05$). Sex interacted with all the isometric strength measures ($P<0.05$) except the plantarflexors at 20 and 30 degrees on both legs. Sex also interacted with all the isokinetic strength measures ($P<0.05$) except all plantarflexor contractions, and faster concentric dorsiflexion bilaterally. Means and SE data are in Appendix 3.

4.4 Discussion

4.4.1 Criterion for inclusion in the faller group

In the principal analysis, fallers were defined as those having had at least one unexplained fall over the past 12 months. This can be justified as follows. First, this criterion has been used in most analogous studies (Lord et al. 1999, Daubney and Culham 1999, Gehlsen and Whaley 1990, Whipple et al. 1987) except two (Skelton et al. 2002, Studenski et al. 1991). Second, the possibility that the single fall was a fall that could occur to anyone and thus not constitute a true unexplained fall was greatly reduced by ensuring that a full history of the fall event was taken. This permitted elimination of falls due to factors such as icy conditions or heavy collisions. Third, the criterion of one fall provided enough contrast between groups to show differences. Moreover, there were few significant correlations between number of falls and strength or power, suggesting that one fall alone indicated that a threshold of frailty had been passed and that once this threshold was passed the risk of further falls was not increased. There were some associations of number of falls with co-activation and asymmetry but these were not highly significant. This suggests that a greater number of falls may be more linked to lifestyle or risk-taking than strength or power. It is also possible that age differences in sensory and central processing function may have contributed, and this will be explained later in the discussion. Finally, an additional analysis using only fallers who fell 3 or more times did not lead to very different results. In general, this frequent faller analysis led to far fewer findings, possibly because of the low number of fallers falling ≥ 3 times (16). Only four variables – asymmetry of power, eccentric plantarflexion strength at $150°.sec^{-1}$, and co-activation in the weak hamstrings isometrically and the weak dorsiflexors concentrically – showed a group difference where no difference had been seen before.

4.4.2 Isometric strength

Across all four muscles and both legs, there was a clear and significant trend for isometric strength to be lower in fallers. However, analyses of individual variables show that this effect was more pronounced in certain muscles and at certain angles.

4.4.2.i Quadriceps isometric strength

This study's results suggest that isometric quadriceps strength is only associated with falling at shorter muscle lengths. All other comparable studies on community dwelling fallers have measured quadriceps isometric strength at 90 degrees flexion only (MacRae et al. 1992, Daubney and Culham 1999, Lord et al. 1999, Skelton et al. 2002, Takazawa et al. 2003, Robinson et al. 2004) which may explain why Daubney and Culham (1999), Skelton et al. (2002), Melzer et al. (2004) and Robinson et al. (2004) did not find differences between fallers and non-fallers. However, other studies have discerned differences in isometric quadriceps strength between fallers and non-fallers at this angle (MacRae et al. 1992, Lord et al. 1992, Lord et al. 1999, Takazawa et al. 2003). In contrast to this study, and others not showing a group difference, these studies used hand-held dynamometers instead of more sophisticated dynamometry to measure strength, which could explain the difference. In addition, the longitudinal nature of the Takazawa et al. (2003) study may have increased sensitivity through the greater power of paired analyses.

The finding of differences only at shorter muscle lengths could be due to an altered length-tension relationship in the fallers towards a peak at shorter muscle lengths than the non-fallers, but this is not evident in Fig. 4.2. This finding could be an extension of differences between young and old, as age related changes in quadriceps strength have also been shown to affect longer muscle lengths less than shorter lengths (Lanza et al. 2003, Fisher et al. 1990). Lanza et al. (2003) suggested that a reason for the greater age decline at shorter quadriceps lengths may be that activities involving more acute knee angles, such as standing from a chair, may be performed regularly, whereas activities where the knee angle is more obtuse, such as climbing stairs, may be performed less frequently. They add that a fear of falling may exaggerate this tendency, which may partially explain these findings.

Test-retest reliability has been shown to be greater for the longer quadriceps lengths (unpublished findings, Appendix 1), so the specific lack of findings at longer muscle lengths are unlikely to be due to greater measurement variability.

4.4.2.ii Hamstring isometric strength

Reduced hamstring isometric strength appears to be associated with falling at all points in the range. Fig. 4.5 shows that the fallers had a shifted length/tension relationship in the weaker leg in relation to the non-fallers. This suggests that the fallers may have optimal filament overlap at around 50° knee flexion (with 90° hip flexion), as opposed to 30° for the non-fallers. In the stronger leg (Fig. 4.4) a similar but less exagerrated trend occurred, with the fallers' strength remaining fairly constant across the range of motion.

MacRae et al. (1992) and Robinson et al. (2004) also observed hamstring isometric strength differences between fallers and non-fallers, at the only angle tested of 90° knee flexion. In contrast, Skelton et al. (2002), Daubney and Culham (1999) and Melzer et al. (2004) found that reduced hamstring isometric strength was not associated with falling, at 90° flexion in the former studies, and at an unspecified angle in the latter. Reasons for these discrepancies are not obvious.

4.4.2.iii Dorsiflexion isometric strength

The results suggest that reduced dorsiflexion strength is only associated with falling at the angles corresponding to the lengthened position, away from plantargrade. This concurs with the findings of Mac Rae et al. (1992), Daubney and Culham (1999), Roma et al. (2001), Takazawa et al. (2003) and Robinson et al. (2004), who all noted an association between dorsiflexion strength in the "neutral" position and falling. Melzer et al. (2004) did not observe any group differences, but they did not specify the angle tested.

The similarity between groups in strength at plantargrade may result from this position being at the far end of range for many of the subjects in both groups. It may therefore be that discomfort at this angle prevented fully motivated efforts, and thus widely varying and confounding results, although no test-retest reliability data is available to support this assertion. However, this does not explain the results seen at 10° plantarflexion, which were within all subjects' comfortable range.

Another possible reason could be a shift in the length-tension relationship in the fallers towards a peak in tension at shorter dorsiflexor lengths. This is not evident in Figs. 4.6 - 4.7, but the lack of an obvious trend may be due to the low number of angles. Such a length-tension shift might occur as a result of reduced plantarflexion range in the fallers. However, there is evidence that reduced plantarflexion range does not occur in fallers (Gehlsen and Whaley 1990) and so this possibility is unlikely.

4.4.2.iv Isometric plantarflexion strength

For the plantarflexors, the highly consistent finding of an association between falling and reduced strength at all angles bilaterally is partially supported by Robinson et al. (2004), who noted a difference in the "mid range" position, which was the only position tested. However at "neutral dorsiflexion", Daubney and Culham (1999) did not observe a group difference and Melzer et al. (2004) noted no group difference at an unspecified angle. Reasons for these discrepancies cannot be explained by methodology.

4.4.3 Concentric strength

There was a consistent and significant trend for concentric strength across all muscles, legs and speeds to be lower in fallers. Given the isometric results, these are unsurprising findings. Since falls usually occur in a dynamic context, dynamic strength should bear a similar or better relationship with falling than isometric strength.

Concentric strength in the plantarflexors was consistently lower in fallers at both speeds. The plantarflexor isometric and concentric results agreed well, and this very consistency may underline the importance of weak plantarflexors as a key factor in falls. Although Skelton et al. (2002) did not note any differences between fallers and non-fallers in concentric plantarflexion strength, these results agree well with those of Studenski et al. (1991), who noted a difference in concentric plantarflexion strength between fallers and non-fallers, in a methodologically similar study to this one.

In contrast, concentric strength in the other three muscle groups did not show significant differences when individual variables were considered, or when variables for each muscle group were averaged within contraction types. This concurs with previous findings for the hamstrings (Skelton et al. 2002, Studenski et al. 1991) and quadriceps (Skelton et al. 2002, Schwender et al. 1997). However, other work has demonstrated deficits in quadriceps (de Rekeneire et al. 2003, Studenski et al. 1991) and dorsiflexor (Skelton et al. 2002, Studenski et al. 1991) concentric strength in fallers. Importantly, de Rekeneire et al. (2003) noted a small but significant difference between groups of around 7% in a large scale study, suggesting that small concentric strength differences in the quadriceps may relate to falls risk, differences that were too small for this relatively small-scale study to detect individually.

One possible reason for differing dorsiflexor results in the Studenski et al. (1991) and Skelton et al. (2002) studies was that fallers were classified as those having had at least 2 or 3 falls respectively, thus suggesting those fallers were frailer. However, when fallers in this study were redefined as those with $\geq$3 falls, the result did not change.

4.4.4 Eccentric strength

This is the first study to show eccentric strength differences between fallers and non-fallers, demonstrating a strong trend for fallers to have lower eccentric strength across all muscles and both legs. However, this tendency was greatest in the plantarflexors, with significant individual variable group differences underlined by averaged differences. This confirms the importance of the association between plantrflexor strength and falling already demonstrated in the other contraction types. In contrast, the hamstrings and dorsiflexors showed only isolated differences between individual variables, whilst the individual quadriceps variables did not differ. In the only other study to measure faller and non-faller eccentric strength in community-dwelling subjects, Skelton et al. (2002) also did not observe differences in eccentric quadriceps strength. Eccentric quadriceps strength therefore seems to be of less importance.

4.4.5 Mechanism of the relationship between strength and falling

The association between falls and isometric, concentric and eccentric strength in the lower limbs presents three possibilities. First, previous falls can cause a loss of leg strength. This appears unlikely in this study. The falls were generally not injurious and so it is unlikely that any direct trauma had a result on strength. Any negative psychological effects may curtail activity levels (McKee et al. 1999, Scaf-Klomp et al.

2003, Alexander 2001) which might impair strength (Gauchard et al. 2003) but activity levels between groups were similar.

Second, leg strength and falls may have a correlative but non-causative relationship, through strength being proportional to the causative factor. Several studies have suggested that strength may not be a causal factor in falls. Lee at al. (1999) measured joint torques during gait in fallers and non-fallers. They found that the fallers had greater torques in most lower limb muscles, except the plantarflexors, and concluded that muscle torques were therefore not a cause of falls. However, this conclusion ignored the fact that the amount of torque used in gait may bear no relationship to any reserve capacity for use in emergency situations such as a postural perturbation (Tang and Woolacott 1998). Ringsberg et al. (1999) concluded that because strength does not correlate with the ability to maintain balance on a balance platform, then strength cannot be a cause of falling (Ringsberg et al. 1999). However, this conclusion failed to recognise that the dynamic righting reactions in response to a potentially fall-inducing pertubation (when the whole body may have high horizontal momentum) are different to the relatively subtle reactions measured on the balance platform (when the whole body may have much lower horizontal momentum).

Wojcik et al. (2001) showed that the actual torques employed in unsuccessful balance reactions in the elderly may not necessarily be smaller than those used in successful balance reactions. This does suggest that the lower strength levels seen in fallers may not be the limiting factor in falling. Pavol et al. (2002) showed that increased strength actually increased the risk of falling for subjects using a particular form of recovery strategy. These subjects also had the fastest walking speed, which was suggested as the causal factor for falling. Similarly, de Rekeneire et al. (2003) showed that the strongest quintile of male older subjects were at a high risk of falls, and greater risk- taking behaviour in these subjects was suggested as the cause. However, it is unlikely that walking speed or risk taking behaviour were primary causes, as young people who have similar gait speed and lifestyle profiles do not fall to the same extent. It is more likely that another factor such as reduced power was responsible. For example, moderate strength with disproportionately low speed, a likely factor in older people who have lost faster units but have remained active and thus retained or even increased (Aniansson et al.1992) strength in their slower units, might enable a good walking pace, but inadequate power to recover from a fall initiated by the faster pace.

This evidence does appear to contradict the final possibility - that reduced isometric, concentric and eccentric strength may be a direct cause of falling. However, strength may be an important causal factor proactively, when the time available for response is not greatly limited.

Adequate clearance over obstacles will prevent tripping, which may be involved in over half of falls in the elderly (Blake et al. 1988). As this may depend on the ability to maintain safely aligned, but not necessarily fast, lower limb trajectories, it may be more strength- than power-dependent. Adequate stance leg plantarflexion and quadriceps strength are required to elevate the whole body over the obstacle (Lamoureux et al. 2003, Begg and Sparrow 2000), which concurs with the findings of lower quadriceps and particularly plantarflexor strength for fallers in this study. In addition, adequate swing leg dorsiflexor (MacRae et al. 1992, Takazawa et al. 2003, Lamoureux et al. 2003) and hamstring strength (MacRae et al. 1992, Lamoureux et al. 2003) is necessary for foot clearance. This relates well to this study's findings of reduced dorsiflexion and particularly hamstring strength in fallers.

Sometimes the trigger for a fall may not be wholly external, but may also result from instability resulting from an inability to control the momentum of body segments, which in turn is caused by muscle weakness (Moxley Scarborough et al. 1999). In this context, avoiding instability requires tonic strength rather than fast movements. There is evidence that during gait, weak plantarflexors in older people may lead to compensatory increases in hip and trunk torque (De Vita and Hortobgayi 2000, Mc Gibbon and Krebs 1999, Mc Gibbon et al. 2001, Lee et al. 1999) which may possibly impair postural stability (McGibbon et al. 2001). Simoneau and Krebs (2000) also showed that fallers had lower plantarflexion moments during gait, and that these were associated with a less steady gait pattern.

It has also been suggested that the deccelerating capacity of eccentric contractions may assist stability (Bellew et al. 1998, Delbaere and Bourgois 2003, Hurley 1995, Lark et al. 2003, Lamoureux et al. 2003). During stair descent, which often leads to falls (Startzell et al. 2000), adequate eccentric plantarflexion strength is believed to be necessary to maintain sufficient joint stiffness for stability (Lark et al. 2003), and joint stiffness is reduced in older people in the stance leg during stepping down (Lark et al. 2003). This agrees well with the findings of weak eccentric plantarflexion for fallers in this study.

Hence stronger subjects may be able to reduce falls by avoiding trips and inducing less self-generated postural instabilities. In other words, strength may prevent a fall from being initiated. However, if a trip or slip does occur, and a subject needs to execute recovery strategies to avoid falling, then speed may become important. It is for these incidents that greater strength alone may not be adequate to avoid a fall.

4.4.6 Lower limb power

The results show that fallers have lower maximal extensor power normalised to body mass in the more powerful leg. This agrees with the findings of Skelton et al. (2002), who found that normalised power in the less powerful leg was associated with falling.

It is surprising is that whilst absolute isometric quadriceps strength showed a bilateral association with falls at the inner range (with normalisation for weight having little effect) leg extensor power only showed a relationship with falls when normalised to body weight and in the more powerful leg. There are four possible reasons for this less pronounced relationship between falls and power. Firstly, the leg extension movement in the power test also involved the hip extensors, the strength of which was not measured. However, hip extensor strength has also been shown to be reduced in fallers (Daubney and Culham 1999) and so this is unlikely to be the sole reason. Secondly, the range of movement in the power test included the quadriceps in its lengthened range, where significant strength differences were not seen. Thirdly, there was a greater ratio of slow to fast concentric strength in the non-fallers, suggesting that in this study the fallers retained faster strength relative to slower strength better than the non-fallers. Whilst this finding is still consistent with reduced power in fallers, it may explain the less pronounced than expected difference. Finally, more strength variables were measured, increasing the chances of significant results. However, despite the less consistent association between power and falls, there was no difference in the index of isometric quadriceps strength / power across the two groups, suggesting that there was a similar age-related loss of both strength and power in both groups.

At a given concentric speed, a greater force represents greater power, and the greater concentric force for the plantarflexors in the non-fallers is perhaps a more convincing demonstration of a power difference between the groups. Importantly, although there was a significant trend for most concentric strength variables to be numerically higher in fallers, significant concentric force differences were not seen for individual variables in the other muscles, which suggests that plantarflexor power is particularly important with regard to falling.

The association between power and falls could be of three types. First, falls could reduce power, but this is unlikely for the same reasons given for falls not causing reduced strength (section 4.4.5). Second, falls and power may have a correlative relationship. This may be the case for falls that can be avoided by high strength independent of speed, as described previously. Third, reduced power may cause falls through a lack of both speed and strength. This is most likely in situations where a trip or slip has already occurred, and the subject needs to respond quickly (Grabiner et al. 1993, Thelen et al. 1996).

Biomechanical studies have supported the latter contention, showing that recovery from a trip requires the rapid contraction of the hamstrings and dorsiflexors in the swing leg to ensure obstacle clearance (Eng et al. 1994, Schillings et al. 2000) and a powerful contraction of the stance leg plantarflexors, hip extensors and quadriceps to elevate the centre of mass to provide extra time to execute a safe swing leg landing (Eng et al. 1994, Pijnappels et al. 2005a, Pijnappels et al. 2005b). A contraction of the stance leg hamstrings also occurs to reduce forward angular momentum before landing (Pijnappels et al. 2005a, Schillings et al. 2000). Pijnappels et al. (2005b) noted that older subjects who fell in response to a trip had a slower rate of increase in muscle activation than younger subjects and those not falling, which underlies the importance of muscle power in successful recovery from a trip. The plantarflexor contraction is required to be particularly powerful (Pijnappels et al. 2005a), which may explain why fallers had lower plantarflexion power in the present study.

Experiments where subjects are released from a forward lean position have demonstrated the need for powerful stance leg plantarflexor activity to reduce anterior rotation of the body at the ankle, and hamstring activity to restrain trunk flexion (Thelen et al. 2000). In the swing leg, dorsiflexion and quadriceps activity provide the torque necessary to execute a long recovery step, and eccentric activity of the dorsiflexors and hamstrings may also help to absorb impact forces on landing (Thelen et al. 2000). Recovery from a forward slip simulated by floor translation in young subjects appears to require powerful activation of the dorsiflexors, rectus femoris and biceps femoris in both the slipping and support limb (Tang et al. 1998). In forward slips induced by an oily floor, the stance leg knee flexors and hip extensors appear more important (Cham et al. 2001). These findings demonstrate the necessity for power in both lower limbs if recovery is to be achieved. It is therefore likely that power below a critical threshold is a cause of falls.

It is important to note, however, that low power may not be the ultimate cause of unexplained falls in healthy people. It is theoretically possible to have high power through very high strength in the presence of very slow contraction speed, which would probably not assist recovery from a trip or slip. As many of the examples above have suggested, it is possible that velocity of movement may be the limiting factor for postural recovery, and so increased power resulting solely from increased strength will not improve recovery. Hence lack of speed, rather than power *per se* may be the final cause. Speed was not directly measured in this study, but future studies should investigate this.

Finally, mention should be made of the implications of any ageing effects on sensory or central processing. One of the assumptions made at the beginning of the study was that subjects would not differ in confounding sensory or central processing deficits as any pathologies involving such sytems were exclusion criteria. However, it is possible that normal ageing effects on these systems may have been present, and that these may have differed between fallers and non-fallers. There is evidence that normal ageing reduces proprioception (Hurley et al. 1998), cutaneous sensation (Quilliam and Ridley 1971, Schmidt et al. 1990), afferent conduction velocity (Hasegawa et al. 1993) and leads to changes in the basal ganglia (Kapellar et al. 2004). The speed and accuracy of afferent information, and the quality of central integration, is vital for an appropriate motor response to be made within the limited time window in which recovery from a postural disturbance can be made. For example, the speed of motor response to a trip may depend upon the quality and speed with which cutaneous information from the foot contacting the object can be sent to the central nervous system and processed. Measurements of the time between contact with an obstacle and an EMG response indicate that older subjects may have slower afferent and processing speeds (Tang et al. 1998), although this has not been shown in other studies (Thelen et al. 2000, Pijnappels et al. 2005b). Further work is required to evaluate differences in response latencies in fallers and non-fallers, but it is possible that such deficits may have contributed to falling in the "faller" subjects. In turn, this suggests that strength and/or power may not be the only causative factors of falling, as suggested by the results in this study. This may explain the inconsistent results with regard to the relationship between different aspects of strength/power and falling. For example, some of the more powerful subjects may have had sensory or central deficits sufficient to lead to increased falls risk. Further work could include measurements such as visual and tactile reaction time to elucidate the influence of such factors.

It is also possible that age-related sensory and integrative deficits could themselves lead to reduced maximal strength and power, thus indicating that such non-motor deficits are possibly the final causes of falling. This could also mean that all or part of the relationship between strength/power and falling could be correlative rather than causal, if sensory deficits were a cause of falls, and caused proportional motor deficits. The mechanism by which sensory deficits could reduce strength or power is likely to be through a loss of feedback leading to reduced voluntary activation, as sensory feedback increases cortico-motoneuronal excitability (Kaelin-Lang et al 2002). However, voluntary activation did not differ between fallers and non-fallers, indicating that differences in strength and power between groups were unlikely to have been significantly influenced by subtle non-pathological sensory deficits. Moreover, there is evidence that the sense of effort (Lafargue et al. 2003) and the ability to perform rapid movements (Rothwell et al. 1982) are largely unaffected by sensory losses.

4.4.7 Association of number of falls with strength and power

Lower limb power, isometric and concentric strength in all 4 muscle groups, and eccentric strength in all but the hamstrings, did not appear to influence the number of falls. This is supported by the fact that strength differences between groups were not noted more frequently when 3 falls was used as the criterion for faller status. Lord et al. (1991) also observed no difference in quadriceps or dorsiflexion strength between multiple and single fallers, although they also did not observe a difference between non-fallers and single fallers.

This implies that although reduced power and strength may be associated with a single fall, as shown in previous sections, further falls are not dependent on this. Hence there appears to be a threshold of strength or power; if this is not reached falls are more likely, but the probability of falling is not increased by being further from this threshold. This may be because a minimum level of strength or power is necessary to avoid a fall, and the likelihood of falling will therefore be the same whether that minimum level is 90% or 10% attained.

The positive correlation of the number of falls with eccentric strength in the hamstrings is surprising. This implies that the greater the hamstring eccentric strength, the greater the number of falls. One explanation could be that people with stiffer hamstrings, and therefore potentially higher eccentric hamstring strength, might be less able to fully extend the knee, and thus find it harder to raise the toes from the ground to avoid obstacles. However, this does not concur with the finding that fallers had lower hamstring eccentric strength than non-fallers. The likelihood is that given the weak significance of these findings, and the large number (84) of correlations performed in this analysis, this result represents a type I error.

4.4.8 Differential effects on contractions and speeds

This is the first study to investigate whether fallers and non-fallers differ in their ratios of isometric, concentric and eccentric strength, and shows that falling is not associated with differences in the relative strength of different contractions. However, fallers appear to have a greater deficit of slow than fast eccentric and concentric strength relative to non-fallers, and this is the first time this has been investigated.

A key assumption in the literature is that fallers may be mechanically slower than non-fallers (Skelton et al. 2002). This may be accomplished by fallers losing more fast muscle mass (Levy et al. 1994) or suffering a greater general slowing of contraction/relaxation than non-fallers. The finding that fallers retain faster eccentric strength better than non-fallers supports this, as slower cross bridge cycling will tend to spare faster eccentric contraction strength (De Ruiter and De Haan 2001, Porter et al. 1995). However, the retention of greater fast than slow concentric strength in fallers relative to non-fallers contravenes this assumption and suggests that fallers are faster. As explained in Chapter 3 (page 63), a greater deficit of concentric force at 50 than 150 $^{\circ}.sec^{-1}$ in fallers compared to non-fallers implies a higher Vmax in the fallers. This is therefore a surprising result, although it would not necessarily contradict the finding that fallers are less powerful, as at moderate velocities (i.e. $150^{\circ}.sec^{-1}$) the fallers are still weaker than the non-fallers. The possibility of a Type I error for this finding is suggested by the lack of findings in the hamstrings and dorsiflexors, and only weak significance in the plantarflexors, but the quadriceps findings were highly significant and the averaged values across muscles and legs were also significant. Further studies are required to confirm these observations.

Assuming that this result stands, it is difficult to explain how a greater deficit of slow than fast eccentric and concentric strength might be causally associated with falling. It is therefore likely that fallers fell despite their relative retention of faster strength because of overall reduced strength and power.

4.4.9 Rectus femoris CSA

The lack of difference between fallers and non-fallers in terms of muscle size suggests that muscle atrophy is not directly linked to falling. This also suggests that the differences in quadriceps strength and power between the groups are not significantly due to atrophy. This is reinforced by the lack of correlation between isometric strength and muscle area.

However, it is important to recognise the drawbacks of ultrasound in assessing CSA of muscle. It is not known if fallers and non-fallers differ in the amount of intramuscular fat or connective tissue. If fallers have greater amounts of such tissues, then this may mask real deficits in the area of pure muscle. The use of ACSA rather than PCSA may also have confounded results; the non-fallers were stronger and so their possibly greater hypertrophy may have led to greater pennation (Kawakami et al. 1993), which will have meant ACSA values were a greater underestimation of PCSA in the non-fallers. Finally, the use of the rectus femoris CSA as an index of overall quadriceps muscle size may have led to a non-valid estimation of whole quadriceps CSA, as rectus femoris CSA may not bear the same proportion to overall quadriceps CSA across both groups.

4.4.10 Index of Force/CSA

This is the first study to compare specific strength (albeit merely an index of specific strength) between fallers and non-fallers, and the results show that fallers have lower F/CSA when trends across variables were considered (although individual F/CSA variable differences were not detected, possibly because of CSA measurement noise). Given the trend for lower quadriceps strength, but the very similar muscle areas in the fallers, this result is as expected. However, the limitations outlined in the previous section concerning the measurement of ACSA in the rectus femoris do call these results into question.

4.4.11 Co-activation

The lack of any differences between fallers and non-fallers in voluntary activation suggests that any differences in F/CSA are due to intrinsic strength differences or levels of co-activation.

The significant trend for mean isometric co-activation variables to be numerically greater in fallers suggests that co-activation may contribute to the decreased absolute and specific isometric strength in fallers. There were no differences between groups for individual isometric co-activation variables, but the significant trend suggests some of these may be false negative results, possibly arising from measurement noise. Moreover, when fallers were re-defined as those having ≥ 3 falls in the previous year, greater hamstring co-contraction was noted in fallers during isometric quadriceps contractions in the weaker leg, suggesting that frequent falls are associated with greater isometric co-activation.

Although there were no overall trends for isokinetic co-activation levels to differ between groups, significantly greater plantarflexor co-activation was observed in the weaker leg during eccentric dorsiflexion in both fallers and frequent fallers. However, plantarflexion co-contraction does not seem to have affected net weak leg eccentric dorsiflexion force in this study, as this did not differ between groups.

In addition, frequent fallers had greater dorsiflexor co-contraction during concentric plantarflexion. This concurred with the lower concentric plantarflexion forces measured in frequent fallers. It also agreed with the highly significant positive correlation between the level of weak dorsiflexor co-activation during maximal plantarflexion concentric contraction and the number of falls, and together these findings suggest that greater co-activation of the dorsiflexors during concentric contractions can increase the risk of multiple falls.

In the only other study to directly compare co-activation in fallers and non-fallers, a delayed onset of dorsiflexion due to prolonged plantarflexor co-contraction during the swing phase of gait was observed (Kemoun et al. 2002). Two studies have also shown that older subjects who were less able to regain postural equilibrium than young subjects after a postural pertubation demonstrated prolonged co-activation in response to a simulated slip or forward fall (Thelen et al. 2000, Tang and Woolacott 1998) but these were not subjects who were reported to have had previous falls.

There are at least three ways in which greater co-contraction could be associated with falls. First, the reduction in net agonist force could reduce the efficacy of postural reactions. For example, the increased co-contraction of dorsiflexors during concentric plantarflexion, and the subsequent reduction in plantarflexor force seen in this study in frequent fallers, could increase the risk of falling by preventing an adequate elevation of the body to enable a safe swing leg landing after a trip (Eng et al. 1994, Pijnappels et al. 2005a). Second, co-contraction could affect the optimal timing of muscle activation and deactivation, such as causing a delay in dorsiflexion during the swing phase that could lead to a trip (Kemoun et al. 2002). It has been suggested that strategies to improve co-ordination, such as Tai Chi, may be effective in reducing such timing errors (Tang and Woollacott 1998a) though there is no direct evidence of this. Finally, greater co-contraction could lead to decreased contraction velocity (Narici et al. 2003), which could affect the success of fall recovery strategies.

4.4.12 Asymmetry

The very significant tendency for isometric strength asymmetry variables to be higher in fallers was accompanied by individual differences for some hamstring and dorsiflexion isometric variables. Eccentric hamstring strength asymmetry was also higher in fallers at $50°.sec^{-1}$. This suggests that asymmetry in slower eccentric hamstrings contractions, isometric strength generally, and dorsiflexion and hamstring isometric strength in particular, is associated with falling. Skelton et al. (2002) did not detect any differences in strength asymmetry between fallers and non-fallers in any muscle group, although eccentric hamstring strength was not measured. These are therefore novel findings.

The positive correlation between faster eccentric dorsiflexion asymmetry and the number of falls suggests that higher eccentric dorsiflexor asymmetry may increase the number of falls. This is in contrast to the findings of Skelton et al. (2002) who did not observe a significant correlation. Caution is required, however, as the association between eccentric dorsiflexion asymmetry and number of falls was a weakly significant result amongst a large number of tests.

The negative correlation seen in the eccentric plantarflexors is surprising, indicating that higher asymmetry reduces the number of falls. It is difficult to account for this finding.

With this study's principal definition of fallers as those suffering one or more falls in the previous year, the results did not suggest that asymmetry of power is associated with falling, in contrast to the findings of Skelton et al. (2002). However, when fallers were defined as those having 3 or more falls in the previous year, the same criterion used by Skelton et al. (2002), fallers' asymmetry of power was found to be greater. Hence this study confirms that frequent fallers have greater power asymmetry.

There are no mechanisms in the literature for a relationship between falls and power or strength asymmetry. One possibility is that asymmetrical forces may lead to instability, particularly when both legs are required to work together, as in a recovery from a fall. However, there is no evidence to support this mechanism. In addition, if asymmetry *per se* causes falling, this would imply that a subject who was asymmetrical but still had a very strong or powerful "weaker" leg would be more prone to falling than a generally weak but symmetrical subject.

Another possibility is suggested by the fact that asymmetry in ageing would be most likely to arise from a weakening of one leg rather than a strengthening of the other. If asymmetry is caused by a weakening of one leg, one would expect to see strong negative correlations between the asymmetry and the strength values in the weaker leg, but not the stronger leg. This is very evident in this study, with a far greater negative correlation between the level of asymmetry and strength in the weaker (Isometric dorsiflexion at 20°: $P<0.001$,$R=-0.49$; Isometric dorsiflexion at 30°: $P<0.001$, $R=-0.51$; Isometric hamstrings at 30°: $P<0.001$; $R=-0.36$; Isometric hamstrings at 60°: $P<0.001$, $R=-0.35$) than stronger leg (Isometric dorsiflexion at 20°: $P=0.04$,$R=-0.20$, Isometric dorsiflexion at 30°: $P=0.06$, $R=-0.19$, Isometric hamstrings at 30°: $P=0.24$, $R=-0.11$; Isometric hamstrings at 60°: $P=0.19$, $R=-0.13$) for all those isometric variables showing a significant difference in isometric asymmetry between groups. In addition, the faster eccentric dorsiflexion asymmetry which correlated with the number of falls was also strongly negatively associated with the strength value of the weak leg ($P=0.008$, $R=-0.37$) but not the strong leg ($P=0.47$, $R=-0.08$). Similarly, the association between power in the less powerful leg and asymmetry was far stronger ($P<0.001$, $R=-0.42$) than that in the more powerful leg ($P=0.0013$, $R=-0.31$). Thus it is possible that subjects with greater asymmetry are those that have the greatest weakness or low power on one side, and the falls are therefore likely to be due to that weakness or low power rather than any pure effect of asymmetry *per se*. That is, the weakness and low power was possibly a causative factor of falls, and the asymmetry was merely correlative.

It should be noted that for the only other asymmetry variable to differ between groups, the slower eccentric hamstring contraction, a weakly significant positive association was seen between asymmetry and strength in the weak leg ($P=0.03$, $R=0.45$) and no association was seen in the strong leg ($p=0.12$, $R=0.33$). This suggests that the weaker the weak leg, the less asymmetry, which opposes the isometric findings. However this was an isolated and not highly significant finding.

Further work is required to establish if asymmetry does have a direct rather than correlative effect. If asymmetry is important in its own right, then this has important implications for rehabilitation, as focussing on regaining symmetry rather than simply improving strength bilaterally may then become important.

4.4.13 Differences between fallers and non-fallers as an extension of age differences

Given that falls become more likely with advancing age (Kenny et al. 2001) it is reasonable to assume that healthy fallers fall because they are physiologically more aged than non-fallers of the same chronological age. Most of the results in this chapter support this assumption, with many of the variables showing age declines (Chapter 3) also being worse in fallers compared to non-fallers. Those variables showing an age effect but not showing a difference between fallers and non-fallers do not contravene this assumption, as this can be explained by the physiological age difference being less than the chronological age gap between the young and older subjects. However there are some variables where differences between young and old are reversed in fallers and non-fallers, and these do conflict with the assumption. For example, the existence of differences between fallers and non-fallers in asymmetry variables that did not differ between young and non-fallers, and specific forces appearing to increase with age, but be lower in fallers relative to non-fallers suggest that factors other than greater physiological age may explain the differences between fallers and non-fallers in this respect.

4.5 Conclusions

Reduced strength and power appear to be strongly associated with falling. Assuming a causal relationship, strengthening interventions should be focussed upon the inner range quadriceps, the entire hamstring and plantarflexor range, the outer range dorsiflexors and the mid range plantarflexors. Eccentric work, particularly for the hamstrings, dorsiflexors and plantarflexors should be included. This may help to reduce falls arising from tripping over obstacles or internally initiated postural instabilities. Training to improve power, particularly in the plantarflexors, should also be carried out to reduce falls due to an inability to recover from a perturbation.

Fallers and non-fallers do not differ in muscle size, or central activation, but fallers have lower specific force and inceased antagonist co-activation. The latter may contribute to falling directly, or via effects on strength and power. Fallers also have more strength asymmetry and frequent fallers additionally have more power asymmetry. However, asymmetry *per se* may not be the cause of falling, as asymmetry appears to relate very strongly to the weakness/low power of the worse leg.

Most of the differences between fallers and non-fallers are consistent with differences observed between young and old, suggesting that fallers are physiologically older than chronologically age-matched non-fallers.

5 Muscle steadiness in young and older subjects

5.1 Introduction

5.1.1 Measurement of steadiness in the young and elderly

During isometric contractions, steadiness has been assessed in most studies by measuring the standard deviation of force fluctuations and normalising to the mean force exerted during that contraction, yielding a co-efficient of variation (CoV) of fluctuations (Galganski et al. 1993, Erim et al. 1999, Burnett et al. 2000, Graves et al. 2000, Laidlaw et al. 2000, Semmler et al. 2000a, Christou and Carlton 2001, Hortobagyi et al 2001, Tracy and Enoka 2002). Such normalisation is normally performed because the standard deviations of fluctuations in isometric contractions are usually noted to increase with the absolute mean force (Galganski et al. 1993, Burnett et al. 2000, Laidlaw et al. 2000, Christou and Carlton 2001, Tracy and Enoka 2002). Hence normalisation permits comparison of steadiness independent of force.

During concentric and eccentric (anisometric) contractions the standard deviations of force (Hortobagyi et al. 2001, Schiffman and Luchies 2001), acceleration (Graves et al. 2000) or position (Burnett et al. 2000, Laidlaw et al. 2000, Tracy and Enoka 2002) have been measured. As force and acceleration are proportional, these measurements can be regarded as equivalent. Positional steadiness measurements do not appear to correlate with force/acceleration steadiness measurements (Tracy and Enoka 2002) and so may measure different aspects of steadiness.

Only Graves et al. (2000) and Burnett et al. (2000) have normalised anisometric results. Laidlaw et al. (2000) found that for anisometric first dorsal interosseous contractions, the respective standard deviations of acceleration, position or force do not vary significantly with the load lifted.

In this thesis, the term *unsteadiness* will generally be used to refer to force, positional or acceleration fluctuations, although it has been used interchangeably with the term *tremor. Tremor* has traditionally been used to describe positional fluctuations such as those occurring in the fingers during position holding against gravity (e.g. Halliday et al. 1999).

5.1.2 Age effects on normalised isometric contraction steadiness

Normalised isometric steadiness has been reported to be less in the elderly than the young at low target forces in the first dorsal interroseous (FDI) (Galganski et al. 1993, Burnett et al. 2000, Laidlaw et al. 2000, Semmler et al. 2000a, Vaillancourt et al. 2003), adductor pollicis (Ranganathan et al. 2001a) and the knee extensors (KE) (Tracy and Enoka 2002).

Some studies have reported no age differences in the FDI (Erim et al. 1999), the KE (Christou and Carlton 2001, Hortobagyi et al 2001, Schiffman and Luchies 2001) and the elbow flexors (Graves et al. 2000). This conflict may result from methodological differences. For example, the results of Erim et al. (1999) may be partially explained by their use of target forces of over 20% MVC, as age differences in steadiness were generally only observed at lower target force levels in studies reporting an age effect (Laidlaw et al. 2000, Semmler et al. 2000a, Ranganathan et al. 2001a, Tracy and Enoka 2002, Galganski et al. 1993) although Burnett et al. (2000) did detect an age difference with target forces as high as 75% MVC. Reasons for the non-significant results in the three knee extensor studies (Christou and Carlton 2001, Hortobagyi et al 2001, Schiffman and Luchies 2001) will be explored in the discussion section. Although Galganski et al. (1993), Laidlaw et al. (2000) and Semmler et al. (2000a) did not consider sex as a possible confounder, it has not been found to affect normalised isometric steadiness in the FDI (Burnett et al. 2000) or KE (Tracy and Enoka 2002), although Ranganathan et al. (2001a) did note a greater decrease in steadiness in elderly women in the adductor pollicis.

In summary, the literature appears to indicate that the elderly may have worse isometric steadiness at low force levels in the FDI, but possibly not in the elbow flexors. For the KE, only one study has shown an age effect, whilst three have not. As will be explained in the discussion section, these three negative studies may not be reliable. Since the function of the KE is likely to be an important factor in falls risk, further studies are required.

5.1.3 Age effects on anisometric contraction steadiness

Some studies have found that both concentric and eccentric contractions are less steady in the elderly in the FDI (Burnett et al. 2000, Laidlaw et al. 2000) the KE (Hortobagyi et al. 2001) and the elbow flexors (Graves et al. 2000). Some of these studies also found that steadiness is worse in eccentric than concentric contractions in the elderly, but not in the young (Graves et al. 2000, Laidlaw et al. 2000) though Tracy and Enoka (2002) did not note a difference between contraction types in either age group. Tracy and Enoka (2002) have suggested that worse steadiness in eccentric contractions in the elderly may be due to their being controlled by unique neural mechanisms (Enoka 1996) and that ageing may preferentially affect this type of neural control.

In contrast, Tracy and Enoka (2002) and Schiffman and Luchies (2001) found that age did not affect anisometric steadiness in the KE. The reasons for this will be examined in the discussion section. In the FDI, Christou et al. (2003a) found that the elderly had greater anisometric steadiness than the young. This study involved an identical methodology to that used in Burnett et al. (2000) and Laidlaw et al. (2000) and so methodological considerations cannot explain the conflicting findings. Interestingly, Christou et al. (2003a) did find that steadiness in the elderly decreased with eccentric contractions, as shown by Graves et al. (2000) and Laidlaw et al. (2000), and with increased movement velocity.

Sex has been reported not to affect normalised anisometric steadiness in the FDI (Burnett et al. 2000). However, it has been found to affect non-normalised eccentric KE steadiness in old subjects and non-normalised concentric KE steadiness in young subjects, with men showing reduced steadiness in both age groups (Tracy and Enoka 2002). Hence Hortobagyi and colleagues' (2001) failure to analyse male and female steadiness data separately, or even to describe the sex make-up of their age groups, may undermine the credibility of their findings.

In summary, the evidence suggests that anisometric steadiness may be worse in the elderly at low target force levels in the FDI and the elbow flexors, although this evidence is inconsistent for the FDI and sparse for the elbow flexors. However, the evidence that there is an age difference for anisometric steadiness in the KE is inconclusive.

5.1.4 Functional steadiness

The non-functional nature of the isometric and anisometric steadiness tests may not adequately reflect steadiness during functional activities and may therefore limit any ability to predict functional performance or falls risk. Muscle steadiness tests conducted in a more functional context may be more sensitive, although this has not yet been studied. There have been studies assessing variability of ground reaction forces during static tandem (Jonsson et al. 2005) and one leg (Jonsson et al. 2004) standing tasks, demonstrating some increases in variability of ground reaction forces with age. However, variability of ground reaction forces do not necessarily directly relate to involuntary variations in muscle force, also possibly resulting from purposeful postural equilibrium responses. There is therefore a need for a study to compare muscle steadiness in young and elderly during functional tasks.

5.1.5 Asymmetry of steadiness

No studies have investigated age-related differences in asymmetry of steadiness. Given the evidence that power asymmetry may be of importance in falling (Skelton et al. 2002) knowledge in this area may be important.

5.1.6 Implications for this study

This review indicates that further studies are required to evaluate the effects of age on lower limb steadiness. Work involving more functional measures is also required.

The hypotheses of this study were that in older people:

1. Lower limb isometric, concentric and eccentric steadiness is decreased

2. Eccentric steadiness decreases with age more than concentric steadiness

3. Lower limb functional steadiness is decreased

5.2 Methods

5.2.1 Subjects

The younger and older non-faller subjects (see chapter 2) were used for this part of the study. Older non-fallers will be referred to as "older subjects" in this chapter.

5.2.2 Steadiness tests

Isometric steadiness

Isometric knee extension steadiness data were obtained using the purpose-built chair previously described (Chapter 3, page 42). Signals from the force transducer were fed into the amplifier, which amplified by a factor of 501. Signals were A-D converted (1401, CED, UK) before being passed into a PC, where signals of force against time were displayed and analysed with Signal 2.13 software. A sampling rate of 1000Hz was used.

The subjects were strapped in the chair, and the back support was adjusted to allow the subjects' posterior knee crease to lie just on the anterior edge of the chair and for their lower back to rest comfortably against the backrest. The order of left/right side testing was randomised and the ankle on the leg to be tested was placed inside the cuff attached to the force transducer. The lower border of the cuff was placed level with the proximal border of the lateral malleolus and the height of the force transducer was adjusted to ensure that the cable ran horizontally. The subjects were asked to pull the cable taut and for all subjects the knee angle was between 85 and 95° of flexion at this point.

The subjects were then asked to perform a minimum of 3 MVCs. If the third MVC was the greatest, then further attempts were made until the force dropped below that of the previous trial. Each MVC lasted 5 seconds with a rest period of 20 seconds between them. Subjects were asked to avoid flexing the hip and were asked to hold on to vertical bars at hand level either side of the seat. They were given verbal encouragement ("Push as hard as you can") and were able to see the force trace. Subjects were advised to breathe in before pushing and to breathe out slowly during each MVC.

Horizontal guide lines equating to 10, 25 and 50% of the best MVC were then inscribed on the screen, and the order of testing was randomised. After a minimum rest period of one minute the subjects were asked to contract to the target level and to maintain this force as steadily as possible for 10 seconds ("Try to keep the force constant so that the force trace is as close to the line as possible"). This was repeated for each submaximal level in the prescribed order with a minimum rest period of one minute between them. A further set of trials was then repeated in the same order. The same procedures were then repeated exactly on the other leg.

The steadiest 6 second portion of each submaximal trace was determined by visual inspection. The standard deviation of amplitude of the 6 second portion was measured by Signal 2.13 software and then divided by the mean amplitude over this portion, giving the CoV of force. The steadier of the two trials at each submaximal level was recorded.

Anisometric steadiness

Anisometric (concentric and eccentric) steadiness data were obtained using the same strength testing chair. Attached to the left hand side of the back of the chair was a pulley system consisting of 3 low friction pulleys, a weight of either 1 kg or 5 kg and a non-extensible cord which could be attached to the subjects' left ankles via a cushioned cuff (Fig. 5.1). The pulley system position was adjusted so that the cord between the anterior and posterior pulleys ran parallel to the floor and was at ankle height. The system could not be adjusted in the transverse plane and so was fixed to a position directly behind the left ankle. Attached to the cord between the anterior and posterior pulleys was an Analog Devices ADXL-201 accelerometer (Analog Devices Ltd, MA, USA) sensitive to changes in velocity in the sagittal plane. Signals from the accelerometer were fed directly into the PC and displayed and analysed using Crossbow Analyse X software version 2.02 (Crossbow technology Ltd, CA, USA) which sampled the data at 220Hz.

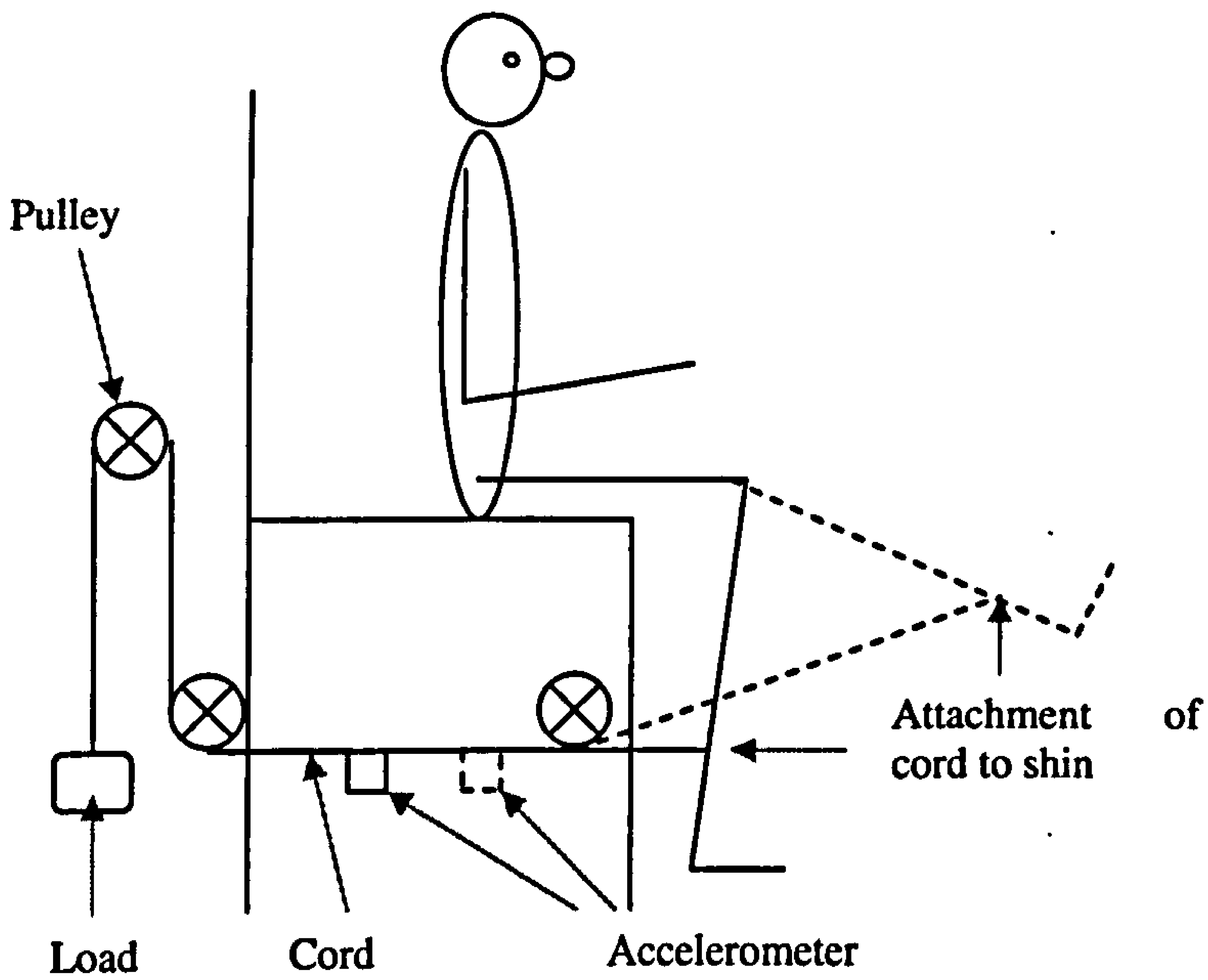


Fig. 5.1 Apparatus for the anisometric steadiness test. Dashed lines indicate positions of leg and accelerometer at end of concentric phase (not to scale).

The subjects were seated in the strength testing chair as above and the left ankle was placed in the cuff, the lower border of which was at the level of the upper border of the lateral malleolus. Only the left side was tested for practical reasons.

Either a 1kg or 5kg weight was then applied in random order to the free end of the cord below the posterior pulley system. The subjects were then instructed to straighten the knee from the starting position of 90° knee flexion to the end position of 30° knee flexion (foot position marked by a stool) over a period of 6 seconds as smoothly as possible at a uniform speed, to hold at the end position for 3 seconds and then to lower the leg back to the starting position over 6 seconds in the same way. Subjects held on to the bars at hand level either side of the seat. They were not required to look at the PC screen during this procedure, but were assisted by the visual aid of a stop-clock with felt-pen markings on the clock face corresponding to the 6 second lift, the 3 second hold and the 6 second lowering. One practice trial was permitted and two measured trials were performed, with a rest of at least 1 minute between them. The procedure was then repeated using the other weight after a rest of at least one minute.

The digital acceleration data were exported to Excel (Microsoft, USA) where they were presented graphically to allow selection of the steadiest 2 seconds of each trial (see Figs. 5.9-5.10). The standard deviation of this portion was then calculated. The steadier of the two trials at each weight was recorded.

Functional steadiness

Steadiness during the functional tests of stepping and sit to stand were obtained using the CODA motion analysis system (Charnwood Dynamics, UK). This system tracks motion of body segments by tracking the 3D location of infra-red emitting markers attached to the subjects' skin.

Markers were placed on each leg at the greater trochanter, the lateral femoral condyle and the lateral malleolus (Fig. 5.2). Two infra-red cameras placed either side of the subject received each marker's infra-red signals and assigned a 3D position co-ordinate to the marker at a frequency of 200Hz. The two cameras (approximately 4m apart) were calibrated to the same co-ordinate reference, allowing signals from a marker to be received by either camera. Location signals from all markers were then fed into a PC and displayed and analysed using the CODA motion analysis software.

Stepping up and down was undertaken on a 20cm high wooden box of 30cm width and 20 cm depth. A 150 cm wooden pole was attached to the back of the box, which provided a means for a subject to maintain balance. Standing up and sitting down was undertaken on an armless chair of 42cm seat height.

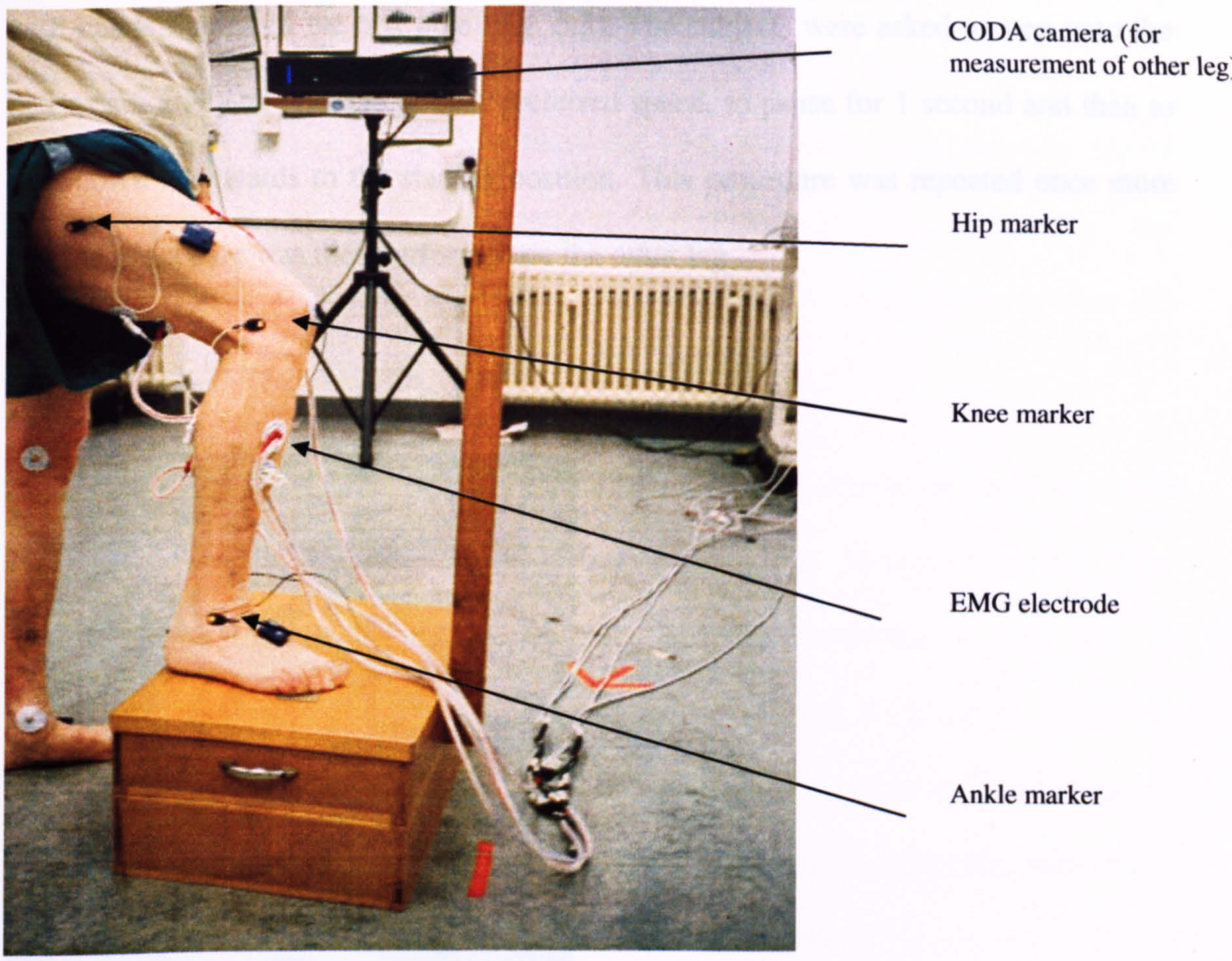


Fig. 5.2. Placement of markers on the hip, knee and ankle.

The subjects sat on the chair in a position from which they felt they could stand up with the greatest ease, and placed their hands in the lap in order to avoid occlusion of the hip markers. Without using their arms to assist the sit-stand movement, they were asked to stand up as smoothly as possible at their preferred speed, to pause for 1 second in the standing position and then to sit down. This procedure was repeated twice.

Stepping procedure

The leg to be tested first was randomised. The subjects placed the test leg on the box with hands lightly on the box pole (Fig. 5.3). The subjects were asked to step onto the box as smoothly as possible at their preferred speed, to pause for 1 second and then to step down backwards to the starting position. This procedure was repeated once more and the procedure was then performed on the other leg

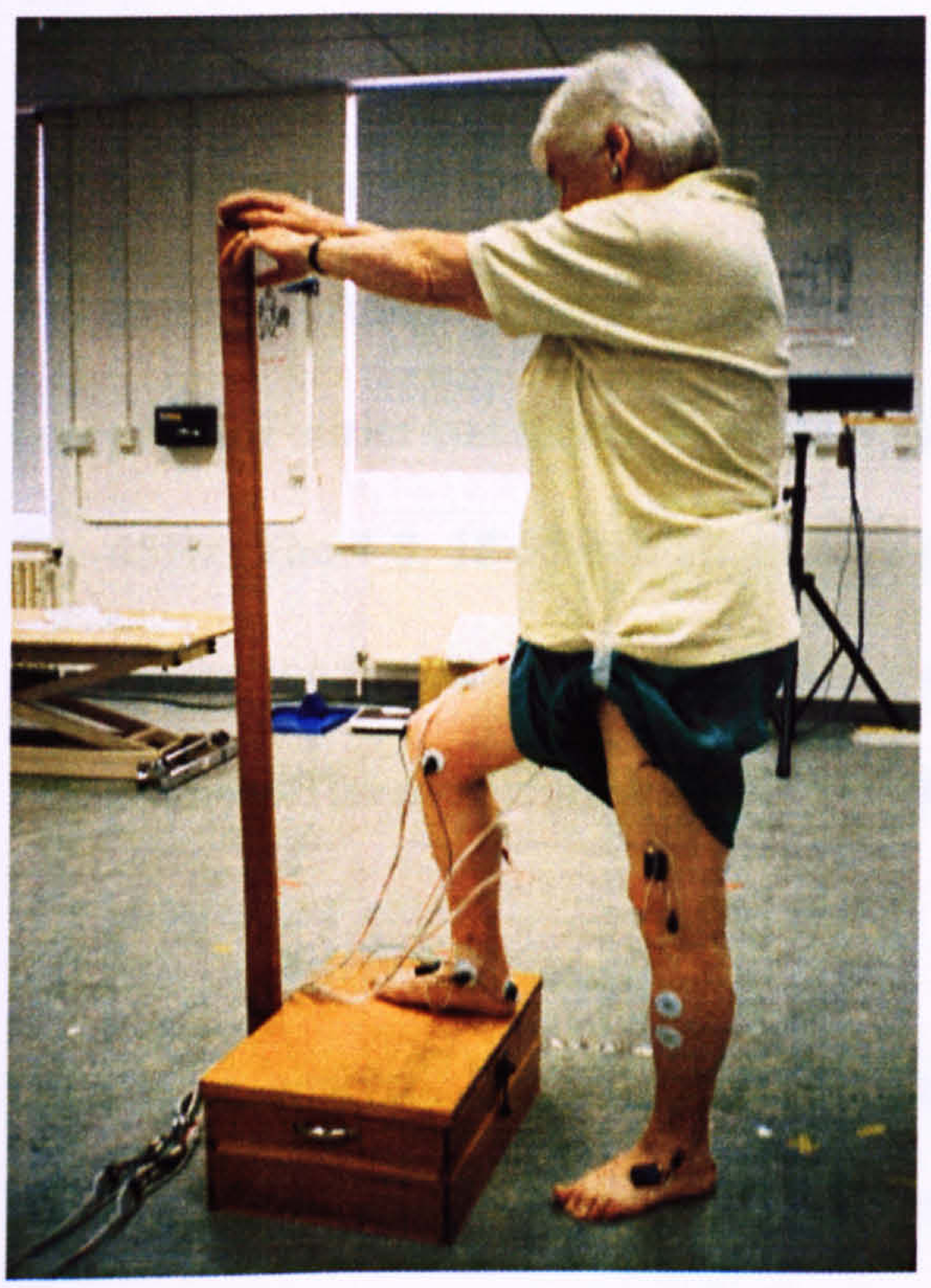


Fig. 5.3. The stepping-up procedure, starting position. Markers are on the lateral side of the right leg.

The marker position data was transformed into knee angle data by the CODA software and the knee angular acceleration over time was derived. Graphs of the knee angle and angular acceleration against time were displayed (see Fig. 5.12). The relatively linear rising portions of the knee angle/time trace equated to the step up or sit-stand up phase, whilst the relatively linear declining portions of the knee angle/time trace equated to the step down or sit-stand down phase. Portions of the angular acceleration graph, which corresponded temporally with these, were selected and the standard deviations of these portions of the angular acceleration traces were calculated as functional measures of steadiness. The steadier of the two trials for each procedure was recorded. SD of acceleration background noise was similar for all groups and tasks at ~2rad sec^{-1} and since it therefore did not threaten the validity of group comparisons it was not eliminated from measurements.

In addition, the angular acceleration data for each manoeuvre on each leg were spectrally analysed. Data were transferred as a data file to Mathcad (Version 2001i Professional, Adept Scientific, UK) where a template (Appendix 4) was used to compute a Fourier transform. Bins were allocated as: 1-4, 4-8, 8-12, 12-18, 18-32 and 32 –45Hz. The 4-8, 18-32 and 32-45 bins were chosen in accordance with known bands of acceleration fluctuation spectra (McAuley et al. 1997), and the other bins were fitted around these.The acceleration power at each bin was calculated.

5.2.3 Statistical analysis

Data were analysed according to the steadiest and less steady leg (except for the asymmetry data) this definition depending on the test used. The method for comparison between groups is described in Chapter 2.

The association between isometric quadriceps strength at 80° knee flexion and steadiness was evaluated using Pearson product correlations for separate young female and older female (non-faller) groups. Male groups were not assessed because of low numbers. Analyses were performed in separate sex/group categories to ensure that effects of age or sex did not distort relationships.

5.3 Results

5.3.1 Baseline group characteristics

The groups did not differ in body mass or corrected height squared (Table 3.1, page 45) but differed significantly in quadriceps isometric strength (Fig. 3.2, page 46). The groups also differed in their numerical sex ratios.

5.3.2 Potential confounding variables

Due to the group differences in sex ratios and the lack of unequivocal evidence that sex differences do not affect steadiness in the lower limbs, sex was chosen as a cofactor for the GLM model. Despite the potential of quadriceps strength to influence steadiness values, this was not used as a covariate in the model as it differed between groups. Correcting for strength would therefore risk correcting for groups, which would nullify results.

During the GLM analyses, comparing steadiness variables across groups with sex as a cofactor, sex interacted with the steadiness variables ($P<0.05$) shown in Table 5.1. Variables not shown did not interact with sex. Group comparisons were corrected for these interactions

Variable	Interacting factor	Direction of interaction
Stand (steadiest leg)	Sex	Women higher values
Stand (less steady leg)	Sex	Women higher values
Sit (steadiest leg)	Sex	Women higher values
Step up (steadiest leg)	Sex	Women higher values
Step up (less steady leg)	Sex	Women higher values
Sit 32-45Hz (steadiest leg)	Sex	Women higher values
Step up 1-4Hz (steadiest leg)	Sex	Women higher values
Step up 4-8Hz (steadiest leg)	Sex	Women higher values
Step up 1-4Hz (less steady leg)	Sex	Women higher values
Step up 1-4Hz (less steady leg)	Sex	Women higher values
Step up 4-8Hz (less steady leg)	Sex	Women higher values
Step down 4-8Hz (less steady leg)	Sex	Women higher values
Average fx steadiness SD variables	Sex	Women higher values
Average fx steadiness 1-4Hz variables	Sex	Women higher values
Average fx steadiness 4-8Hz variables	Sex	Women higher values
Average fx steadiness 8-12Hz variables	Sex	Women higher values
Average fx steadiness 12-18Hz variables	Sex	Women higher values

Table 5.1 Variables interacting with sex and the direction of the interaction. fx = functional, SD = standard deviation.

5.3.3 Age differences in non-normalised isometric steadiness

The young group had higher levels of absolute force fluctuations at 50 and 25% MVC in the steadier leg (Fig. 5.4). Otherwise there were no significant differences between groups.

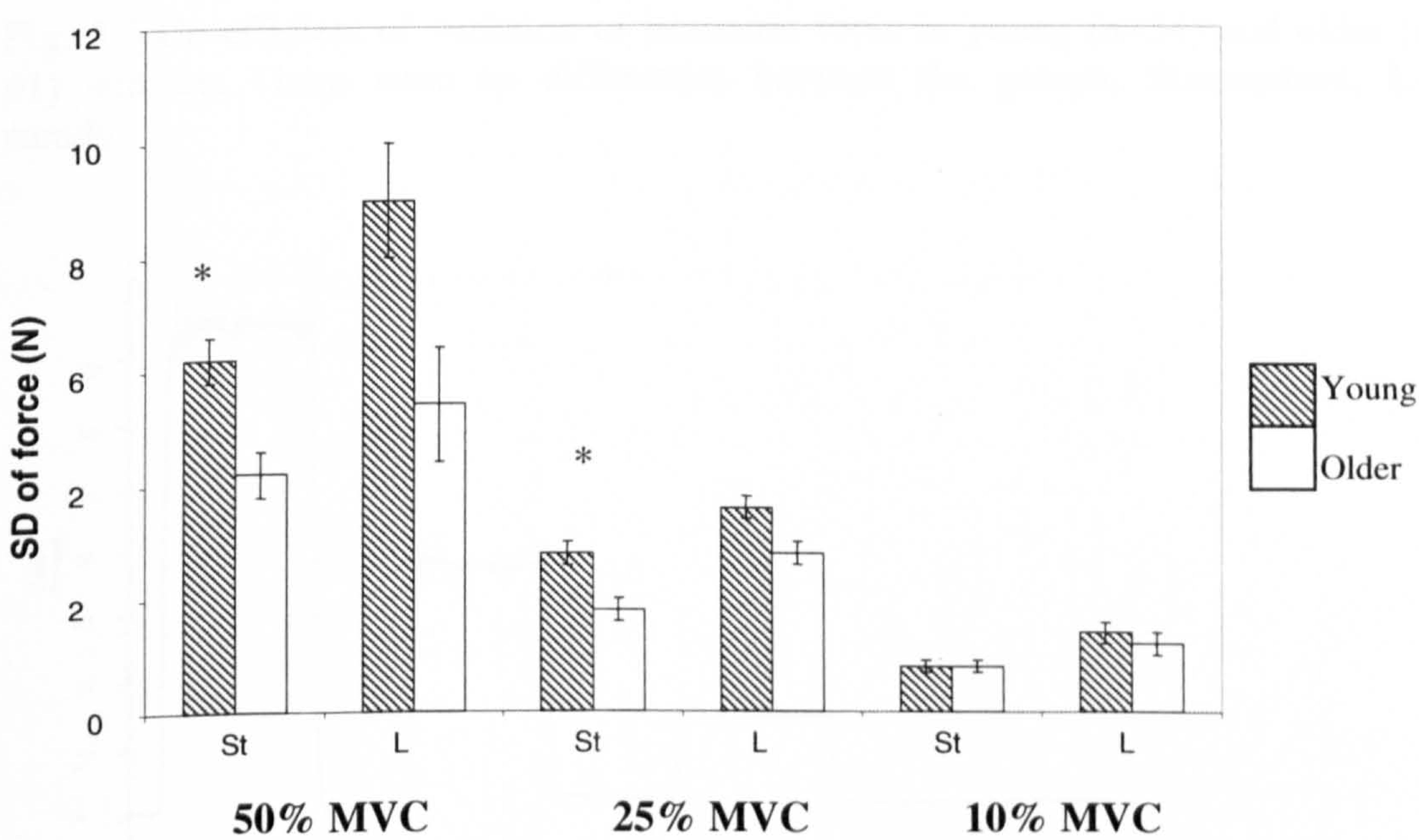


Fig. 5.4 Standard deviation of isometric force for young (n=34) and older (n=39-41) subjects. * P<0.01. St=steadiest leg, L=less steady leg.

5.3.4 Age differences in normalised isometric steadiness

The young and elderly subjects did not differ in normalised isometric steadiness at any of the three submaximal levels (Fig. 5.5). When the bilateral measurements across all contraction intensities were averaged there were also no differences between groups (Young CoV: 1.3±0.1, Older: 1.4±0.1). Figs. 5.6-5.7 show typical force traces for steadier and less steady subjects.

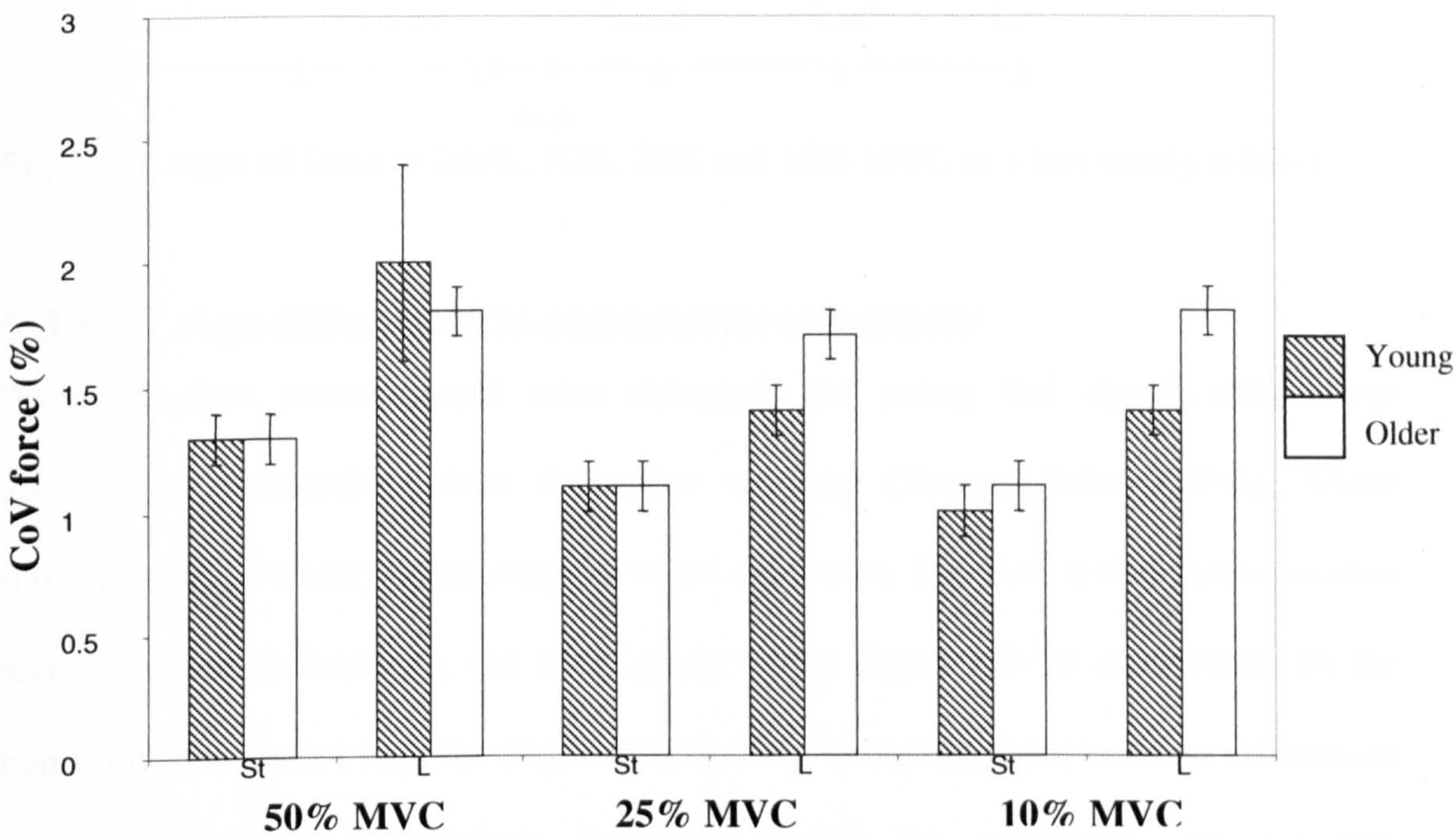


Fig. 5.5 Co-efficient of variation of isometric force in young (n=34) and older (n=39-41) subjects There were no differences between the groups. St=steadiest, L=least steady.

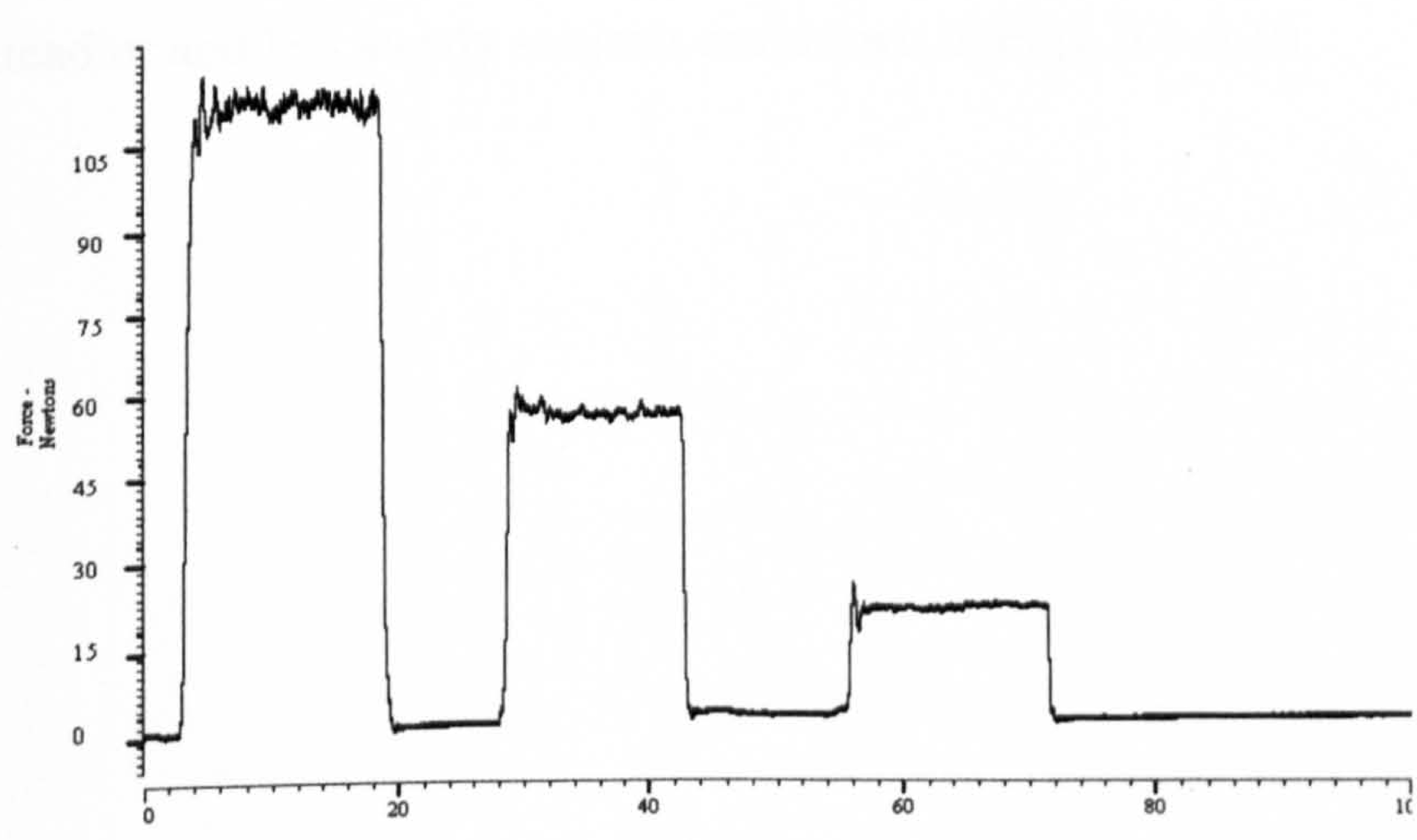


Fig. 5.6. A typical trace of force at 50, 25% and 10% MVC in a steadier subject.

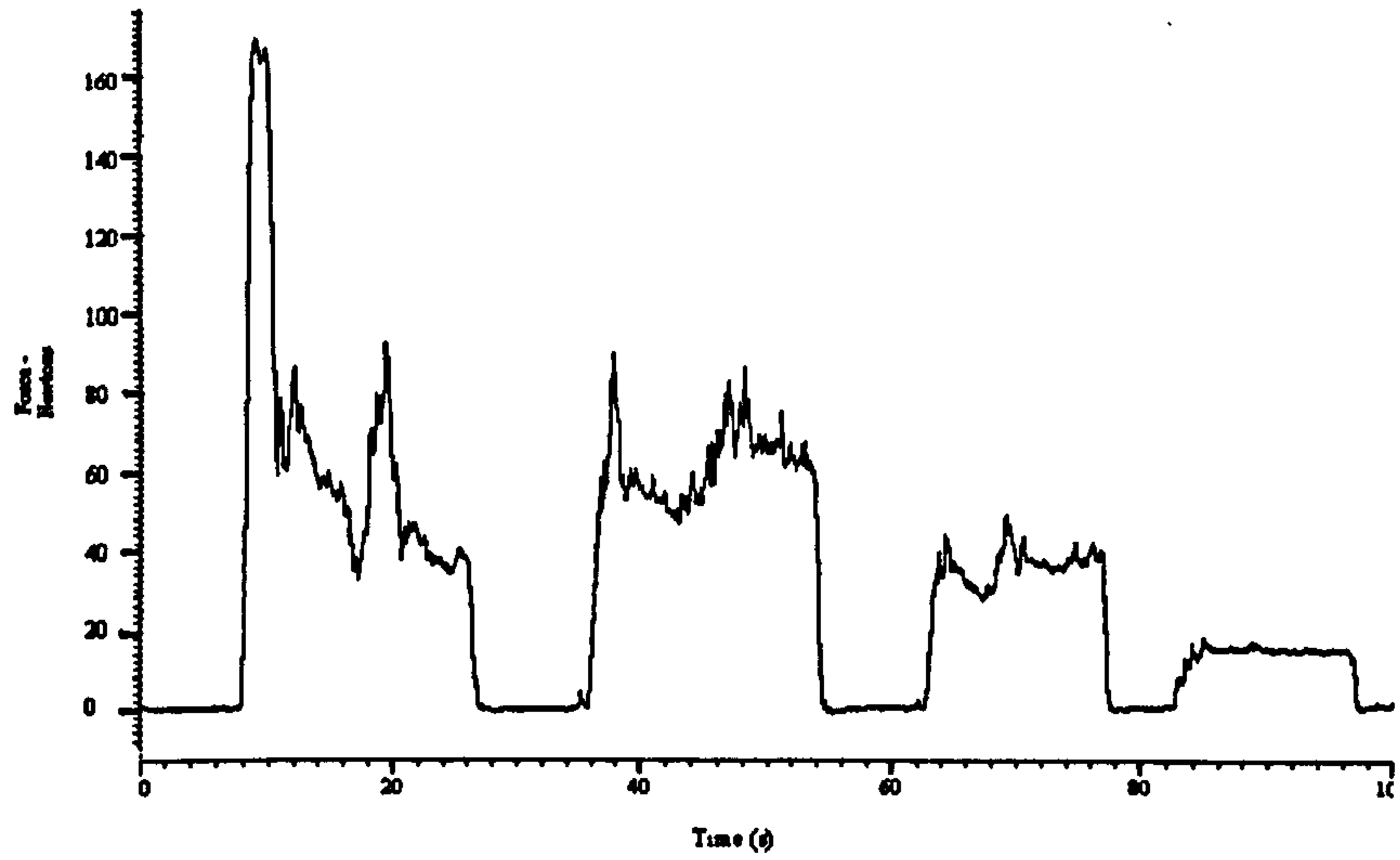


Fig. 5.7 A trace of force at 100%, 50%, 25% and 10% MVC in a less steady subject.

5.3.5 Age differences in anisometric steadiness

When all four measurements were averaged, the young had significantly lower acceleration fluctuations than the older subjects (Young: 0.019±0.002g, Older 0.025±0.002g, $P<0.05$) confirming the trend evident in Fig. 5.8. When measurements were analysed individually, the older group had a higher SD of acceleration in the concentric movement with the 5kg load (Fig. 5.8). However, there were no differences between groups in anisometric steadiness during the concentric and eccentric movements with the 1kg load, or the eccentric movement with the 5kg load. Traces of acceleration against time for the concentric and eccentric phases of the movements in steadier and less steady subjects are shown in Figs. 5.9-5.10.

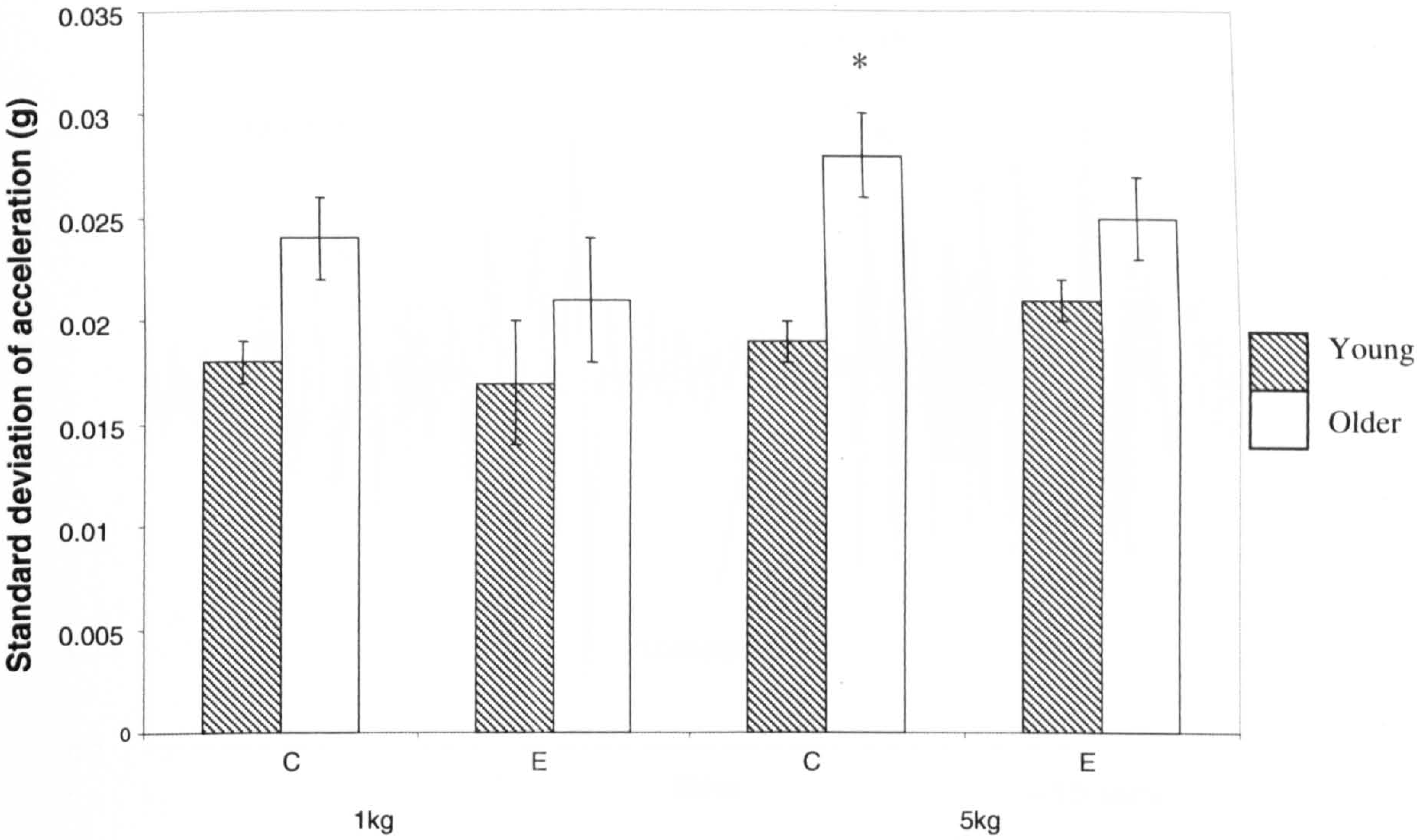


Fig. 5.8 Standard deviation of anisometric steadiness in young (n=23) and older (n=19) subjects. *P<0.01. C=concentric, E=eccentric.

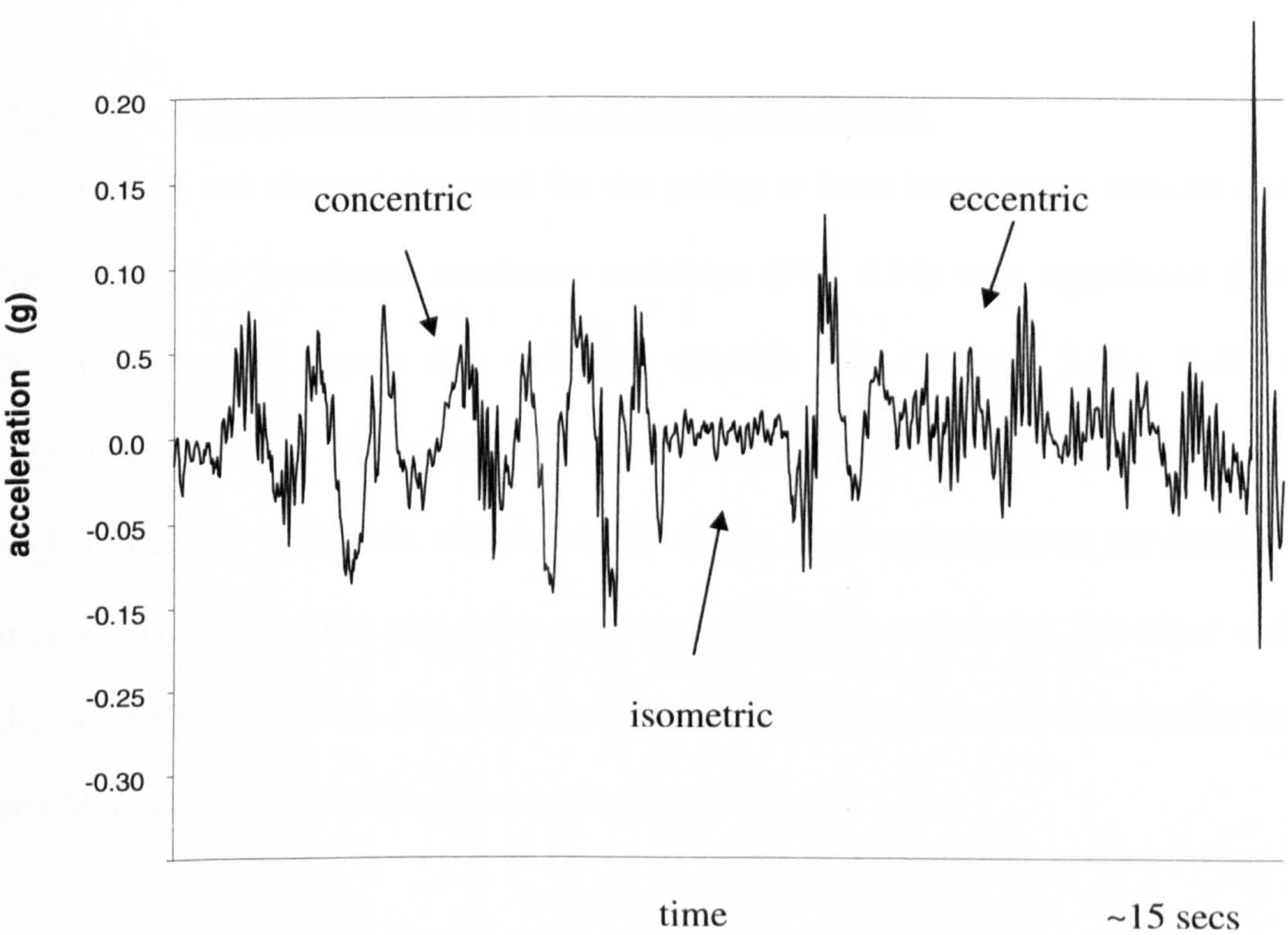


Fig. 5.9. Trace of acceleration (g) against time in a steadier subject with a 1kg load. Only the central 2 second portion of the concentric and eccentric phases was analysed.

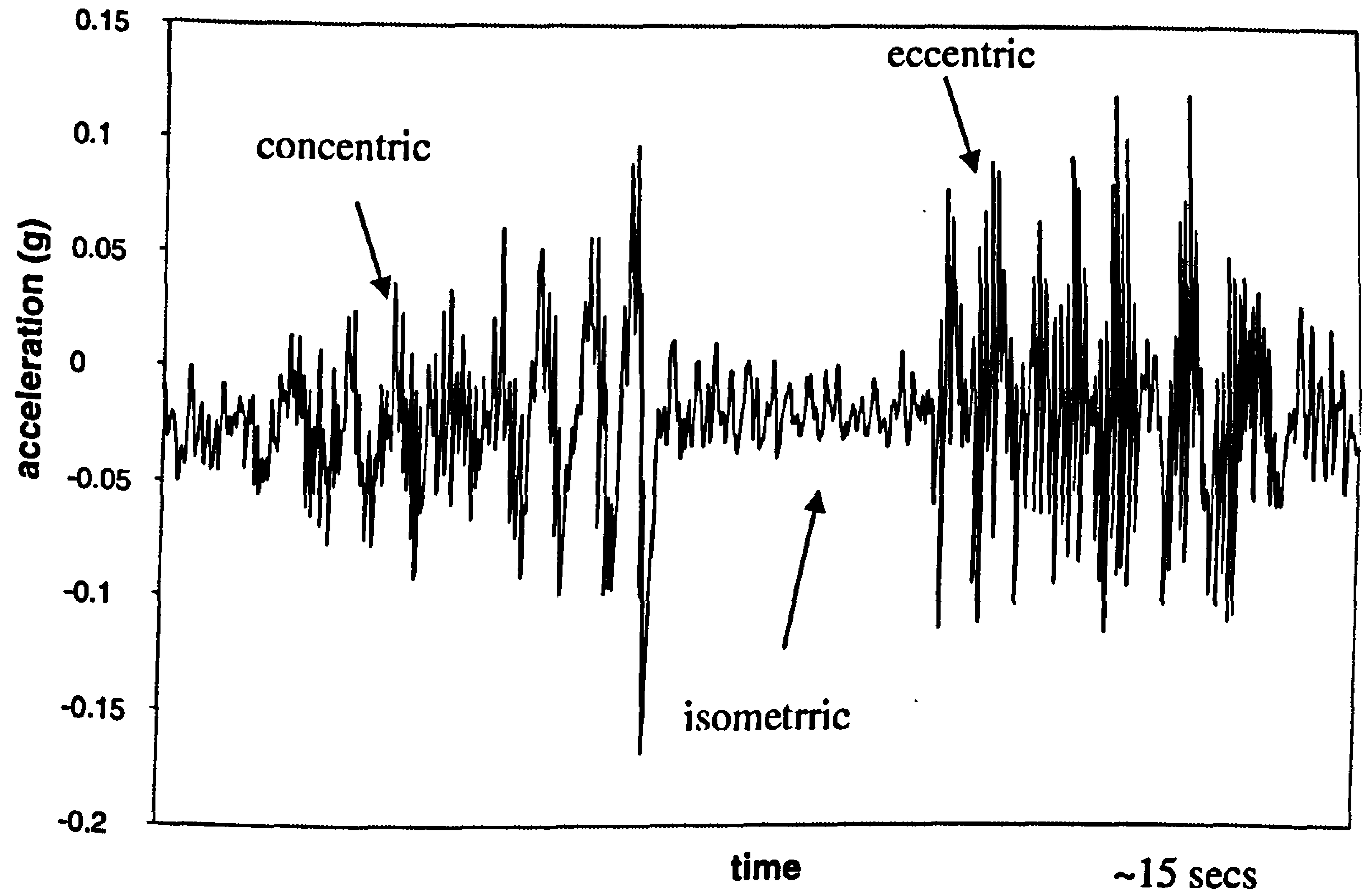


Fig. 5.10. Trace of acceleration (g) against time in a less steady subject with a 1 kg load.

5.3.6 Age differences in functional steadiness

A Binomial test showed the trend for the young to have lower mean numerical values for 8/8 of the functional steadiness variables (Fig. 5.11) was significant (P<0.01). Results averaged across the variables confirm this (Young: 5.45$\pm$ 0.49, Older 7.68$\pm$0.49, P<0.01). When variables were analysed individually, the older group had higher standard deviations of acceleration in the stand manouvre on the steadiest and less steady legs and the step down manouvre on the less steady leg, but other variables did not differ (Fig. 5.11). Fig. 5.12 shows a trace of angular acceleration against load for an older and younger subject during stepping down

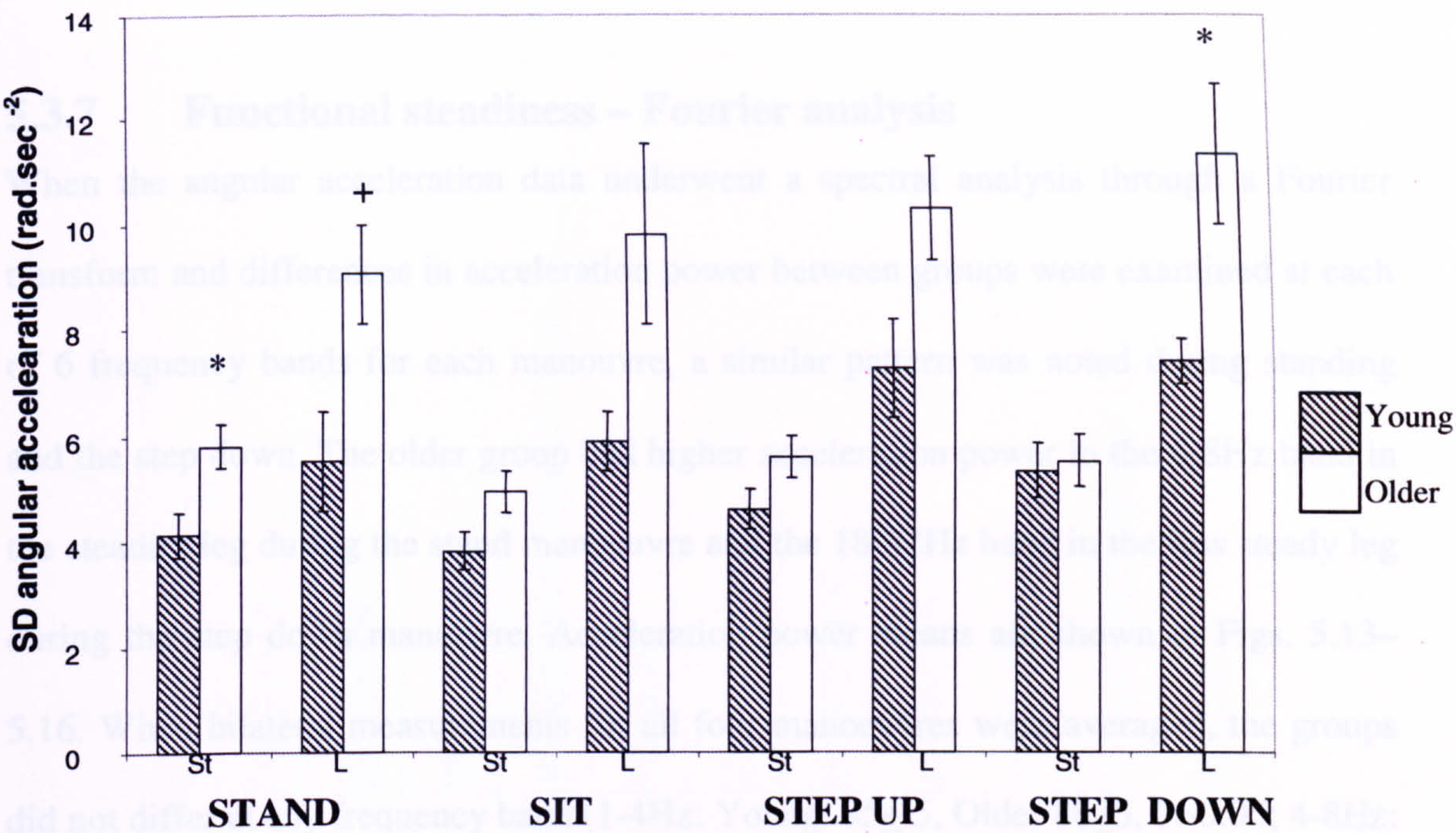


Fig. 5.11. Standard deviation of knee angular acceleration during functional movements in young (n=40) and older (n=38) subjects. *P<0.01. +P<0.05 St=steadiest leg, L= less steady leg.

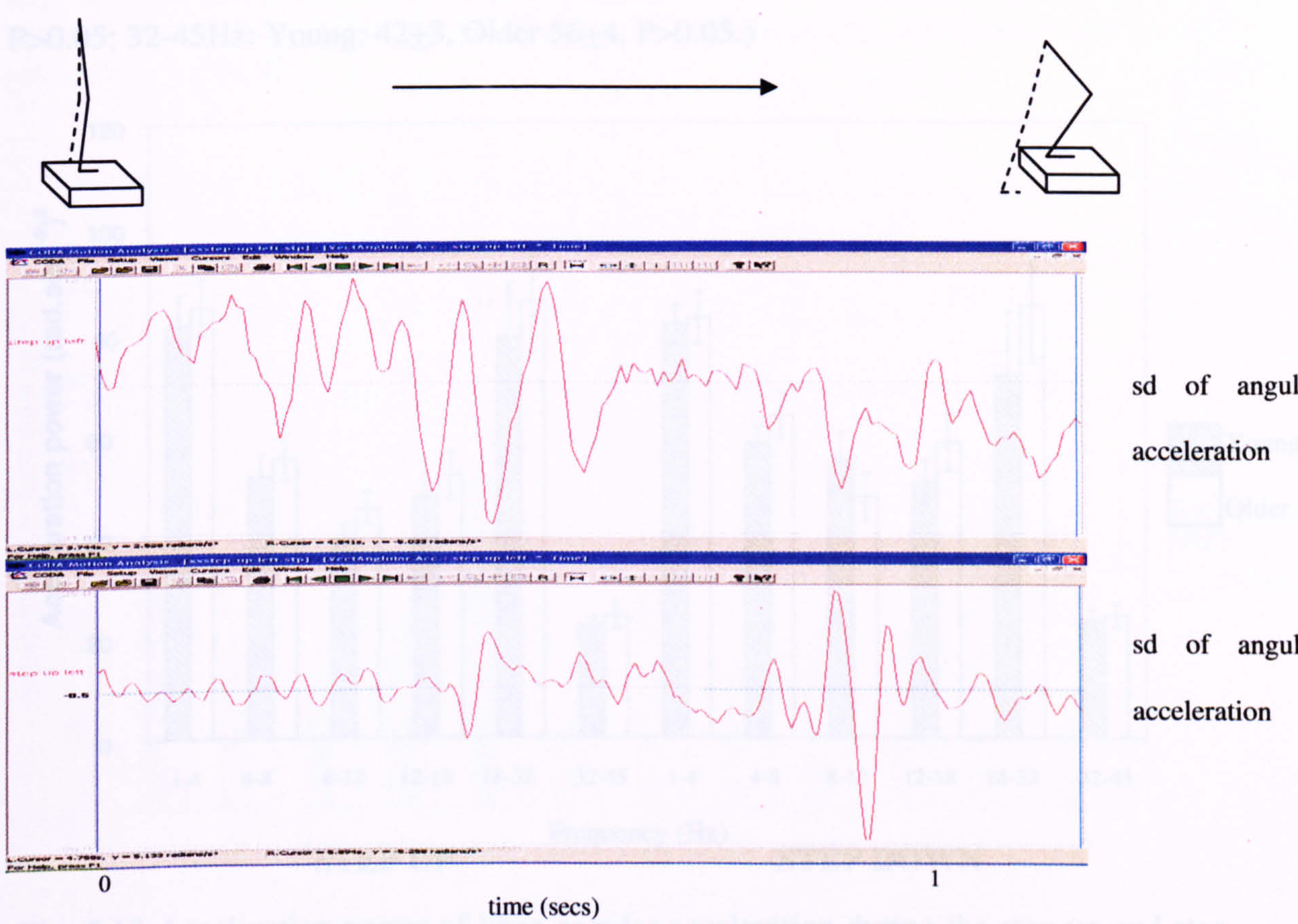


Fig. 5.12 Traces of angular acceleration against time for an older (upper trace) and younger (lower trace) subject during the backward step down. The stick figures at the top indicate the positions of the stance leg (unbroken lines) at the beginning and end of the movement.

5.3.7 Functional steadiness – Fourier analysis

When the angular acceleration data underwent a spectral analysis through a Fourier transform and differences in acceleration power between groups were examined at each of 6 frequency bands for each manouvre, a similar pattern was noted during standing and the step down. The older group had higher acceleration power in the 4-8Hz band in the steadier leg during the stand manoeuvre and the 18-32Hz band in the less steady leg during the step down manouvre. Acceleration power means are shown in Figs. 5.13–5.16. When bilateral measurements for all four manoeuvres were averaged, the groups did not differ at any frequency band (1-4Hz: Young: 85± 5, Older 88±5, P>0.05; 4-8Hz: Young: 68±5, Older 82±5, P>0.05; 8-12Hz: Young: 67±6, Older 76±6, P>0.05; 12-18Hz: Young: 79±11, Older 112±11, P>0.05; 18-32Hz: Young: 140±11, Older 195±18, P>0.05; 32-45Hz: Young: 42±3, Older 56±4, P>0.05.)

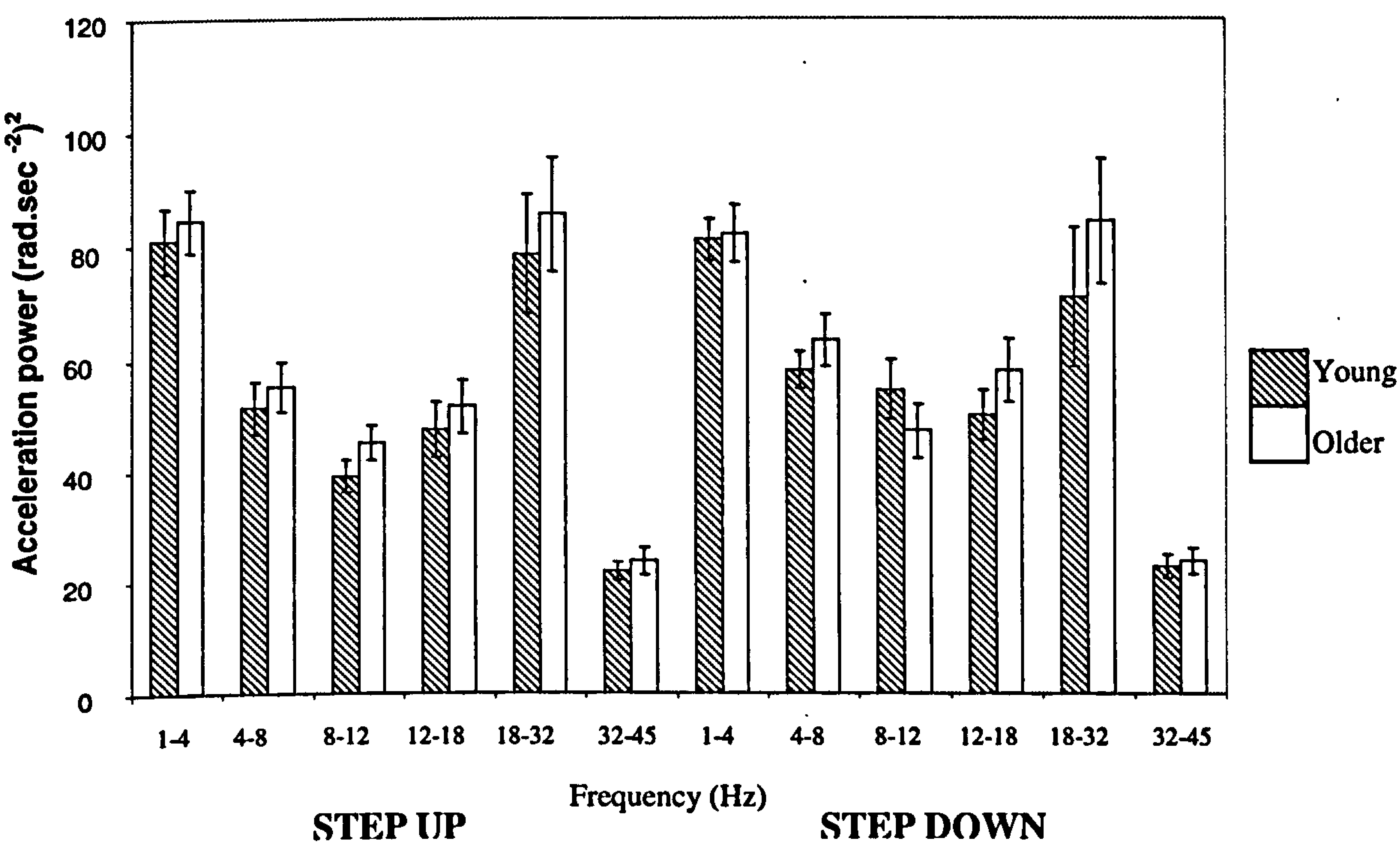


Fig. 5.13 Acceleration power of knee angular acceleration during the step up and step down manoeuvres on the steadier leg at 6 frequency bands in young (n=40) and older (n=38) subjects. There were no differences between groups.

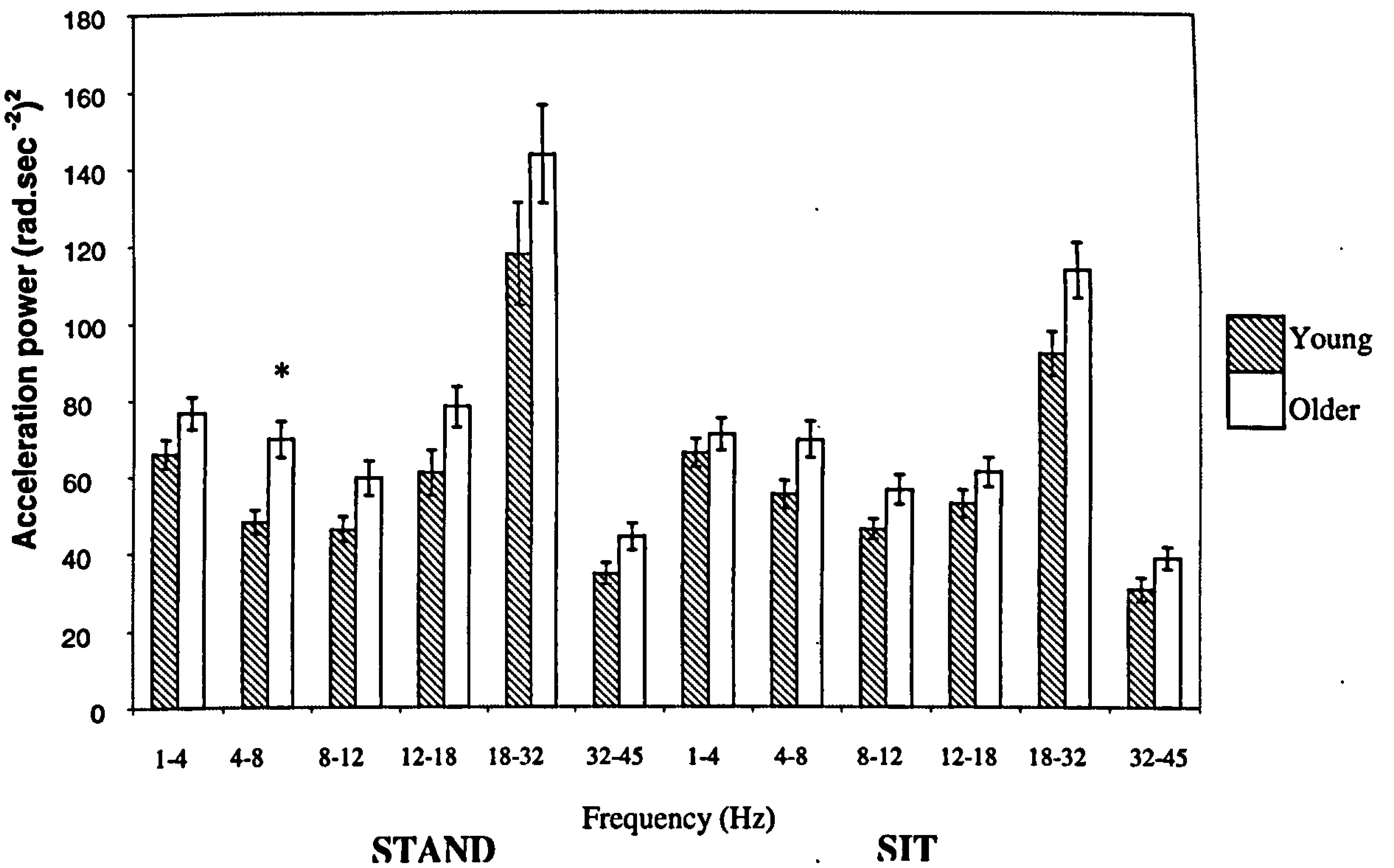


Fig. 5.14 Acceleration power of knee angular acceleration during the stand and sit manoeuvres on the steadier leg at 6 frequency bands in young (n=40) and older (n=38) subjects. * P<0.01

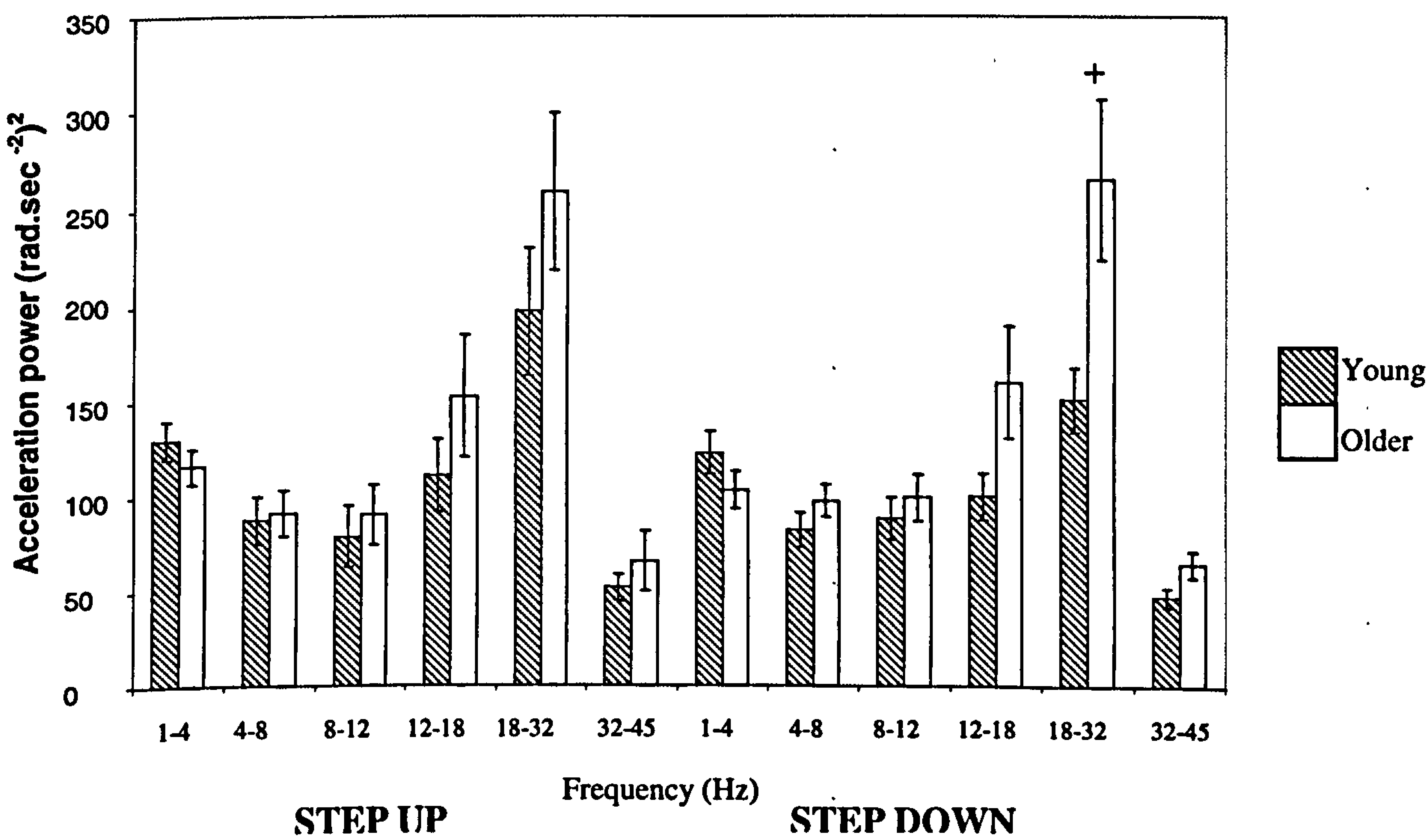


Fig. 5.15 Acceleration power of knee angular acceleration during the step up and step down manoeuvres on the less steady leg at 6 frequency bands in young (n=40) and older (n=38) subjects. + P<0.05.

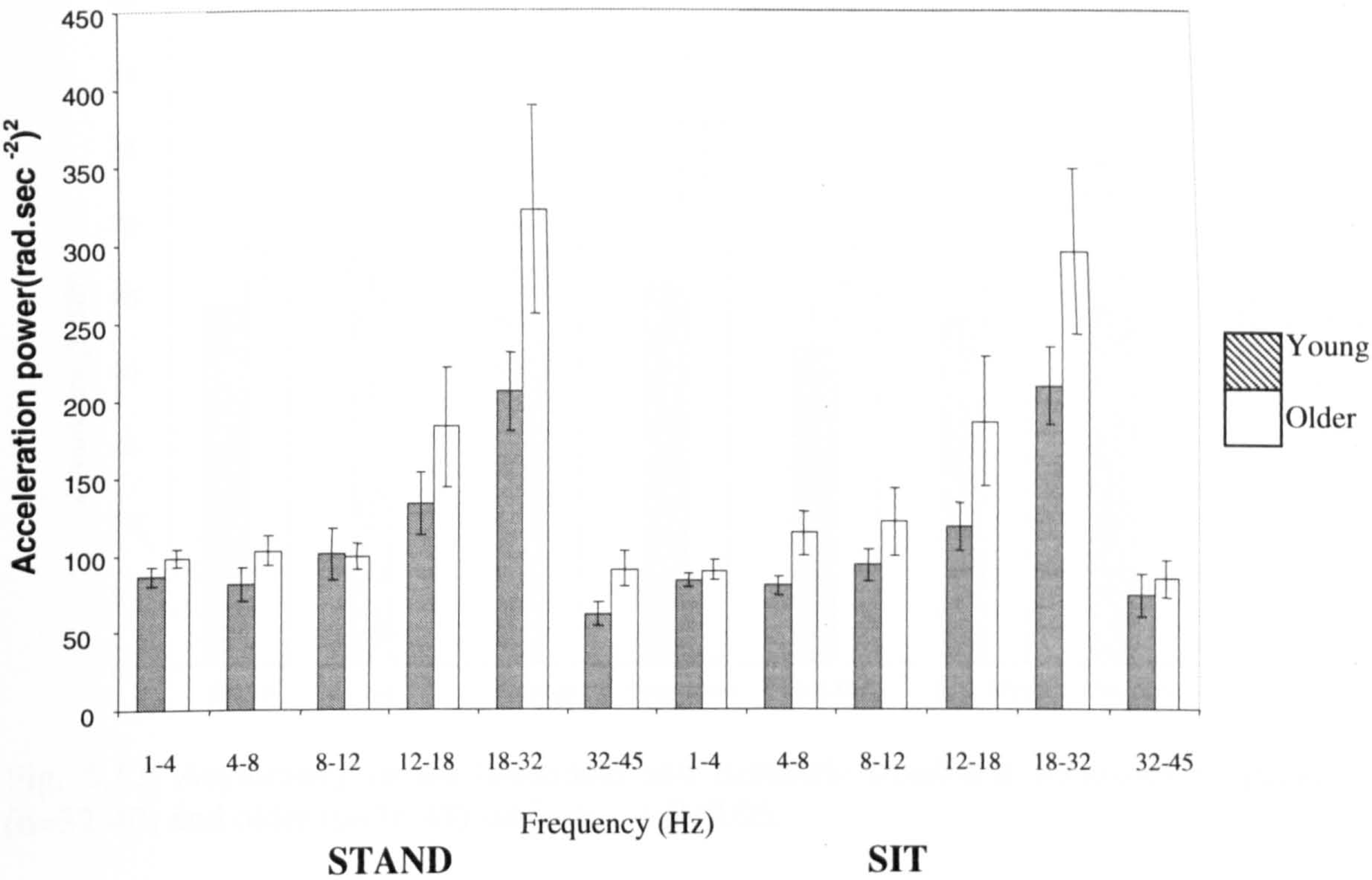


Fig. 5.16 Acceleration power of knee angular acceleration during the stand and sit manoeuvres on the less steady leg at 6 frequency bands in young (n=40) and older (n=38) subjects. There were no differences between groups.

5.3.8 Asymmetry of steadiness

A Binomial test showed that there was a significant trend for asymmetry of steadiness to be greater in the older subjects (P<0.01). When variables were analysed individually, the older subjects had greater asymmetry in the step down manoeuvre only (Fig. 5.17).

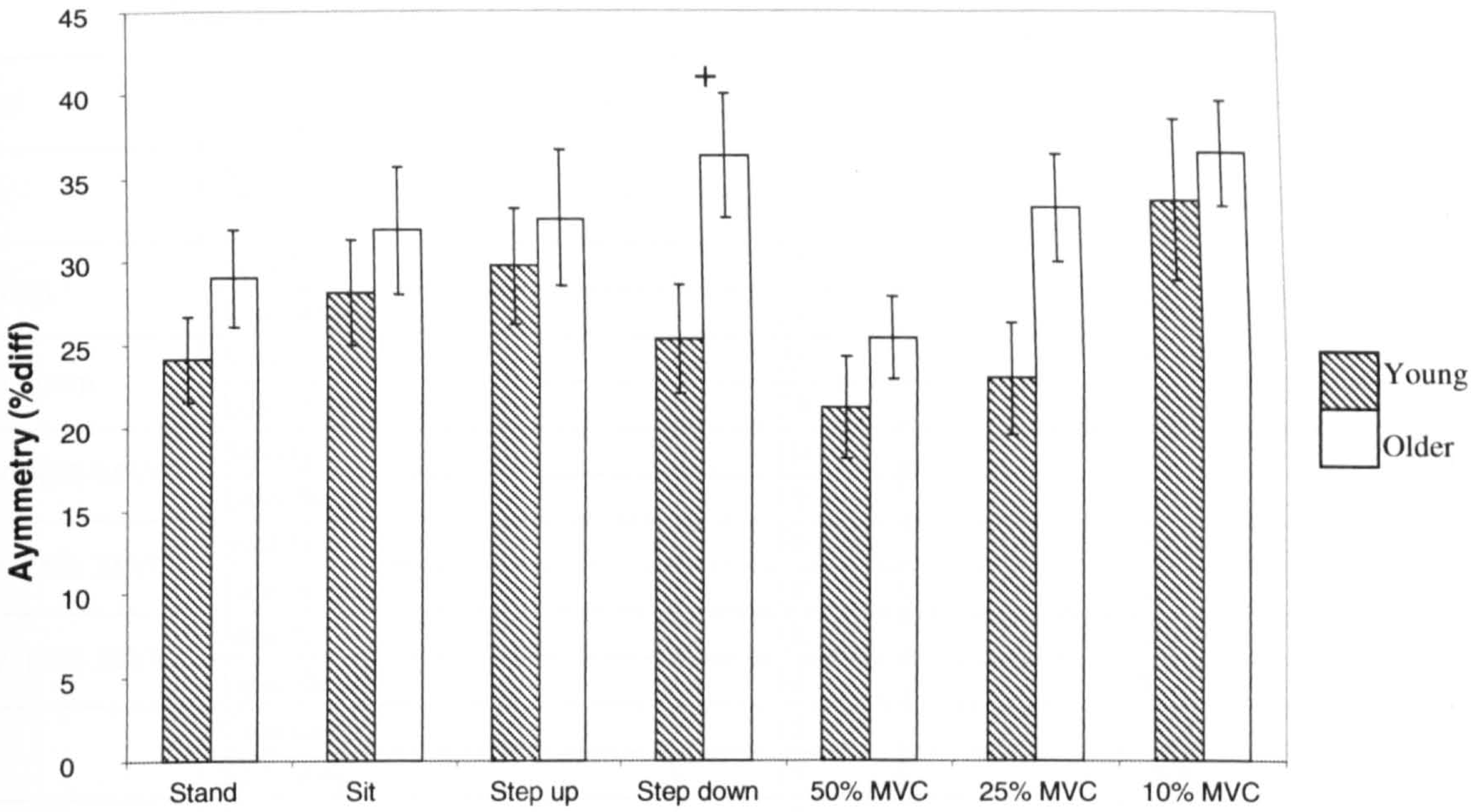


Fig. 5.17. Asymmetry in the functional and isometric steadiness measures in young (n=32-40) and older (n=38-43) subjects. + P<0.05.

5.3.9 Interaction with strength

Young women showed a significant negative correlation (P<0.05) between peak isometric quadriceps strength and the SD of knee angular acceleration during sitting in the less steady leg, and positive correlations between peak isometric quadriceps strength and the CoV of force during the 50%MVC contraction in the steadiest leg, the 25%MVC contraction in the steadiest leg, and the 10%MVC contraction in the less steady legs (Table 5.2). Though it may initially seem inappropriate to correlate low force steadiness with peak strength and power, the aim was to assess any relationship between being steady and being strong/powerful.

Steadiness variable		n	Correlation between steadiness and strength in corresponding leg
Stand	Steady	22	-0.15
	Less steady	22	-0.29
Sit	Steady	22	-0.03
	Less steady	22	-0.50+
Step up	Steady	23	0.02
	Less steady	23	-0.15
Step down	Steady	23	0.18
	Less steady	23	-0.11
CoV 50% MVC	Steady	19	0.57*
	Less Steady	19	0.18
CoV 25% MVC	Steady	18	0.55+
	Less Steady	18	0.37
CoV 10% MVC	Steady	18	0.41
	Less Steady	18	0.58*
1kg	Concentric	13	-0.32
	Eccentric	13	-0.02
5kg	Concentric	12	-0.02
	Eccentric	12	-0.09

Table 5.2 Association between quadriceps strength and steadiness in young women. +P<0.05, *P<0.01.

Older women showed a significant negative association between peak isometric quadriceps strength and the force CoV at 10% MVC in the less steady legs (P<0.05) (Table 5.3).

Steadiness variable		n	Correlation between steadiness and strength in corresponding leg
Stand	Steady	27	0.14
	Less steady	27	-0.05
Sit	Steady	27	-0.10
	Less steady	27	0.01
Step up	Steady	24	0.19
	Less steady	24	0.05
Step down	Steady	24	-0.17
	Less steady	24	-0.19
CoV 50% MVC	Steady	25	0.17
	Less Steady	25	0.14
CoV 25% MVC	Steady	25	-0.07
	Less Steady	25	-0.28
CoV 10% MVC	Steady	26	-0.39
	Less Steady	26	-0.40+
1kg	Concentric	14	-0.29
	Eccentric	14	-0.17
5kg	Concentric	14	0.50
	Eccentric	14	-0.06

Table 5.3 Association between quadriceps strength and steadiness in older women. +P<0.05

5.4 Discussion

5.4.1 Isometric steadiness

5.4.1.i Age effects on the standard deviation of force

This study showed that younger subjects had higher standard deviations of force (i.e. were less steady) than older subjects at the highest force intensity when no normalisation was carried out. This partially concurs with the findings of Tracy and Enoka (2002) who noted that the young showed greater standard deviations of knee extension force than the elderly at 5, 10 and 50% MVC, but not at 2% MVC. Christou and Carlton (2001) also found that the young had greater standard deviations of force in the knee extensors, but that these were more apparent at lower force levels.

Importantly, when steadiness was measured as a co-efficient of variation (CoV) in the present study this difference between groups disappeared. Christou and Carlton (2002) noted the same phenomenon. Tracy and Enoka (2002) have shown an even more stark effect, with the finding that the young were steadier when assessed by CoV. In the hand muscles, similar conflicts have also been found between normalised and non-normalised results, with non-significant age differences in SD co-existing with significantly greater CoVs for the elderly (Galganski et al. 1993, Laidlaw et al. 2000). This reversal is easily explained with an example. At a fixed percentage of maximal target force, assume a young subject has a force SD of 4N, whilst an older person has a lower force SD of 3N. If the young person has an absolute force of 200N at that target level and the older person has an absolute force of 100N, then the CoV of the younger person will be 2% whilst that of the older person will now be higher at 3%. The reversal therefore results from the fact that the proportional difference in absolute force was greater than the proportional differences in SDs.

The conclusions therefore depend upon whether steadiness is normalised, and it is important to be clear about the form of measurement that is used. More importantly, it is vital to consider which is the more valid measure in physiological and functional terms. The argument is complicated by the fact that the submaximal target force levels can also be absolute or normalised. There are therefore four combinations of measurement: 1) SD and absolute target force, 2) SD and relative target force, 3) CoV and absolute force and 4) CoV and relative target force.

1) SD and absolute force

In many situations the elderly are required to perform actions that require the same absolute force as a young person. For example, stepping down will require the same force for both age groups if body mass is equal. Hence measuring absolute fluctuations at an absolute force is a good way of assessing the level of force fluctuations that would be seen in a functional situation. However, in order to establish the effects of age *per se* on steadiness, it is important to control for strength. This method does not allow for such control, as when using a fixed force the weaker older subjects will have to work at a higher percentage of their maximum. The older subjects may thus use different motor unit types and/or different levels of rate coding, which may both possibly influence force fluctuations.

2) SD and relative target force

All studies except Hortobagyi et al. (2001) have used relative target levels rather than absolute target levels for isometric steadiness testing, as this avoids comparison of subjects of different strengths working at very different percentages of their maximum.

However, this may mean that stronger subjects systematically show greater unsteadiness if relative target levels (e.g. 50% of maximum force) are used. This is because standard deviations rise with absolute force level (Galganski et al. 1993, Burnett et al. 2000, Laidlaw et al. 2000, Christou and Carlton 2001, Tracy and Enoka 2002) and the absolute force level at 50% maximum will be higher in a stronger person.

3) CoV and absolute target force

If the absolute target force is constant for both groups, then the ratio of CoVs between the age groups will equal the ratio of standard deviations between the groups. Therefore this method equates to using SD and absolute force.

4) CoV and relative target force

This method has the advantage that steadiness at selected target forces can be assessed independently of strength. The use of normalised force fluctuations also avoids bias due to differing motor unit type recruitment at different levels of strength. However, it fails to recognise that in the functional context forces are rarely adapted to the strength of the individual.

In summary, two methods appear to have benefits: SD with absolute target force and CoV with relative target force. The former is pragmatic, whilst the latter has the capacity to control for some strength effects. The majority of similar studies have used the latter method and comparative concerns provide good justification for its use in this study. In addition, the potential bias created by having different subjects utilising different types of motor units is significant. Hence isometric steadiness will be judged by the CoV at relative target forces, and from this point any references to isometric steadiness will relate to normalised isometric steadiness unless stated otherwise.

5.4.1.ii Normalised isometric steadiness

The lack of an age difference in normalised isometric steadiness of the knee extensors in this study concurs with the findings of Schiffman and Luchies (2001), Christou and Carlton (2001) and Hortobagyi et al. (2001). However, these three studies used different methodology to that used in this study, and so do not necessarily support this study's findings.

Schiffman and Luchies (2001) did not normalise force fluctuations. The non-normalised standard deviations of force tend to rise with absolute force levels in the knee extensors (Tracy and Enoka 2002) and this may explain why the young had numerically larger standard deviations. Christou and Carlton's (2002) findings may be partially attributable their use of a very fit and active group of elderly subjects (Tracy and Enoka 2002). Training studies have indicated that trained elderly FDI muscle has less isometric force variability (Laidlaw et al. 1999) and so such subjects might be expected to be steadier, although isometric quadriceps steadiness does not seem to change with training (Tracy et al. 2004).

The findings of Hortobagyi et al. (2001) are also not directly comparable to those reported here, as they used a fixed isometric force of 25N for both age groups, despite the groups differing in strength. At 25N, the young group were working at 6% MVC, whilst the elderly group were working at 12% MVC. Standard deviations of force were used as measurements of steadiness, but since both groups were producing the same absolute mean force, their standard deviations of force can be regarded as equivalent to co-efficients of variation. As the co-efficient of variation of isometric force fluctuations is generally agreed to *decrease* as the percentage target force level *increases* in both young and old (Galganski et al. 1993, Burnett et al. 2000, Semmler et al. 2000a, Christou and Carlton, 2001) the elderly group might be expected to be exerting a lower

CoV at 12% MVC than they might if they were at the same target level (6%) as the young. This might explain the lack of any significant difference between the age groups.

Both Schiffman and Luchies (2001) and Christou and Carlton (2002) used a different form of feedback to this and all of the other studies on isometric steadiness. The use of direct visual feedback from a force output trace has been the standard method of allowing subjects to regulate their isometric force output (Galganski et al. 1993, Burnett et al. 2000, Laidlaw et al. 2000, Semmler et al. 2000a, Tracy and Enoka 2002). In contrast Schiffman and Luchies (2000) required subjects to keep the force level within a 6.1Nm band around the specified 20 or 60% force level. The only visual feedback given were coloured lights if the force was above or below the band. Thus subjects were not given direct feedback of their actual force levels. As a fixed bandwidth was used regardless of the absolute target force level, the weaker elderly subjects had a relatively larger bandwidth (Tracy and Enoka 2002). This may have affected differences observed between the age-groups. Christou and Carlton (2001) provided no visual feedback at all during the measured portion of the submaximal contractions. A horizontal line was provided at the correct position on the monitor to allow each subject to achieve the correct level, but after 5 seconds the monitor was covered and the subjects had to rely on proprioceptive feedback.

It is unclear if visual feedback has a confounding effect on isometric steadiness values. A meaningful test of steadiness must involve sensory feedback as it is one of the many influences upon movement control, but visual feedback of force output is not the form of sensory feedback used during normal movements, and there is evidence that visual feedback can confound results. For example, it can induce wrist and finger oscillations at 2-3Hz (Kakuda et al. 1999), whilst De Serres et al (2000) have found that increasing the gain of force output display (that the subject uses as feedback) decreases steadiness, which indicates that enhanced visual feedback reduces steadiness. In contrast, Schmied et al. (2000) did not observe a relationship between force display gain and steadiness. A small pilot study of the effects on steadiness of keeping the monitor covered after the force level had been established was carried out in this study (Appendix 5). Although the study suggested that visual feedback improves steadiness at lower intensities in both young and older subjects, this appeared to be due to the force dropping in the absence of visual feedback as the contraction continued, which artifactually increased the CoV values. Although such force data could be usefully analysed for fluctuations after detrending, this would mean that steadiness at discrete contraction intensities could not be assessed. For this reason, and to permit comparison with other studies using visual feedback, and because a large number of subjects had already been tested using visual feedback, it was decided to continue to use visual feedback in the methodology.

The results of this study are in conflict with a similar study on the knee extensors (Tracy and Enoka 2002) where the elderly were found to have greater normalised force fluctuations at 2, 5 and 10% MVC. Subject differences may account for part of this difference. In this study, the elderly group was made up of only non-fallers, whilst in the Tracy and Enoka (2002) study there is no record that fallers were not included. There is evidence from the present study that fallers may have lower isometric steadiness than age-matched non-fallers and so it is possible that the lack of fallers in the elderly group contributed to the difference in findings between studies. There were several methodological differences as well. Tracy and Enoka (2002) measured force fluctuation variation over a longer period (8 versus 6 seconds). In the present study the analysis window was restricted to 6 seconds as it was found that the elderly were less able than the young to maintain a steady force level for more than 6 seconds. Christou and Carlton (2001) have also found that force levels begin to drop after 7.5 seconds of submaximal isometric contraction although they did not report an age difference in this respect. Hence the longer windows in the Tracy and Enoka (2002) study may also have contributed to their finding of an age difference.

Tracy and Enoka's differing results may also be due to their use of lower target levels than this study, as age differences have been noted mainly at lower submaximal levels (Semmler et al. 2000a, Ranganathan et al. 2001a, Laidlaw et al. 2000, Galganski et al. 1993) though this does not explain their differing findings at 10% MVC. Finally, Tracy and Enoka (2002) did not specify whether they analysed limbs according to left/right, steadier/less steady or strong/weak.

The results of this study also conflict with studies on other muscles such as the FDI (Galganski et al. 1993, Burnett et al. 2000, Laidlaw et al. 2000, Semmler et al. 2000a, Vaillancourt et al. 2003) and the adductor pollicis (Ranganathan et al. 2001a). As there are few methodological differences, conflicting outcomes are more likely to be due to the different muscle groups studied. The two main differences between the knee extensors and the small hand muscles are as follows.

First, the knee extensors are made up principally of four large quadriceps muscles, whereas the FDI and AP are small single muscles. It has been suggested that age-related changes to individual motor units, such as increased variability of action potential discharge (see Chapter 7), may only lead to changes in steadiness in small muscles with few motor units (Tracy and Enoka 2002, Graves et al. 2000). This may be because force fluctuations due to discharge variability are more likely to cancel out if there are many active units. The variability of action potential discharge may play a large role in age-effects on steadiness in smaller muscles (Laidlaw et al. 2000, Graves et al. 2000, Tracy and Enoka 2002) and if this role is diminished in larger muscles age differences in steadiness may be lower.

Second, the knee extensors and small hand muscles differ in their relative reliance on rate coding and recruitment to control force (Graves et al. 2000). Small hand muscles tend to use recruitment at low intensities to increase force and by 50% MVC all motor units have been recruited (Graves et al. 2000). In larger muscles, however, full recruitment is not achieved until 85% MVC (Graves et al. 2000). This means that at low forces the hand muscles will be using relatively larger motor units than in the larger muscles, which may still be recruiting smaller motor units at a high frequency (Graves et al. 2000). Since smaller motor units are believed to lead to lower force fluctuations (Galganski et al. 1993) the lower force fluctuations in larger muscles may make differentiation between groups more difficult.

Despite the normalisation of force fluctuations to the absolute load, maximum quadriceps strength correlated significantly with some isometric steadiness values. This association was different between young and old, with greater strength being strongly associated with greater normalised force fluctuations in the young but lower normalised force fluctuations in the older subjects. Although physiological ageing effects may be responsible for this, it is difficult to formulate a mechanism. Nevertheless, this tendency may partially explain the lack of differences seen between young and old in isometric steadiness, given that the young group were stronger.

Another reason for the lack of findings could be the test-retest reliability (Appendix 6). The Least Significant Difference (LSD) value of around 0.007 (percent), and the CoV of measurements of around 15-20% for all isometric measures, are greater than the mean differences observed between groups. This high noise to signal ratio could have obscured any small but real differences. However, although reliability is not reported for other studies using similar methods, the variability of measures as shown by SD were lower in this study than in Tracy and Enoka et al. (2002): the SD in this study for CoV at 50% MVC was 0.39 compared to 0.75 in Tracy and Enoka (2002). Although this difference may partially relate to differences in biological variation of the samples rather than mere differences in repeatability of measures, this does indicate that this measure is inherently variable.

In conclusion, the three knee extensor studies with similar findings do not support this study's findings, due to their very different methodologies. The conflicting knee extensor study (Tracy and Enoka 2002) also had small methodological flaws which may weaken its results. The evidence from this study shows that a group of healthy elderly that does not include any fallers has similar isometric steadiness to a younger group.

5.4.2 Anisometric steadiness

5.4.2.i Age effects on anisometric steadiness in the lower limbs

The worse steadiness in the elderly in the 5kg concentric knee extensor contraction, and the overall worse steadiness when averaged over the four contractions, suggests that anisometric steadiness is greater for older subjects during movement. The lack of correlation between the anisometric measures and strength suggests this difference was not a result of strength differences, and so may represent an intrinsic difference in steadiness.

This result partially agrees with the findings of Hortobagyi et al. (2001), who found that concentric and eccentric contractions were less steady in the elderly. As in this study, Hortobagyi et al. (2001) used fixed loads and non-normalised steadiness measures. This may partially explain the similarity of their concentric results. As previously described, the preferred measurement involves using normalised target forces, as failure to normalise target force may mean that the different age groups are working at different percentages of maximum and so possibly using different strategies to increase force and different sizes of motor units. This may therefore not allow a fair comparison of steadiness *per se*. In this study, the decision to use fixed target levels was dictated by the wish to use a system analogous to the functional steadiness measurements (which also used fixed loads, given that age groups were similar in body mass) to allow a comparison of the two dynamic methods. In retrospect, the potential benefits of this decision probably do not outweigh the drawbacks and so this can be seen as a potential limitation.

Hortobagyi et al. (2001) measured force variability on an isokinetic dynamometer, with the subject using visual feedback to try to keep the force constant. This meant that subjects had to alter activation levels to compensate for the effects of the length/tension relationship through the tested range of motion, which may have provided an extra challenge to participants. In this study the force was constant and the subject was merely asked to maintain a constant velocity. It is therefore possible that the steadiness tasks differed in difficulty. Because of the differing types of sensory feedback it is also possible they may have involved different sensory and motor pathways and therefore involved measurement of different abilities. Caution must therefore be exercised in comparing results.

The results from this study also concur with studies on upper limb muscle groups (Burnett et al. 2000, Laidlaw et al. 2000, Graves et al. 2000) but these agreements cannot be used to strongly support the findings from this study, as previously discussed in section 4.4.1.ii.

In contrast, the results from this study conflict with the findings of Tracy and Enoka (2002) and Schiffman and Luchies (2001). Tracy and Enoka (2002) did not detect any age difference during concentric contractions but noted that the older men and older women were *more* steady (as measured by standard deviations) than their younger counterparts during eccentric contractions at 10 and 5% respectively. Schiffman and Luchies (2001) also did not note any age effect for concentric contractions. There were also conflicting findings in the study of Christou et al. (2003a) but these were on hand muscles and this prohibits meaningful comparison.

There were some important methodological reasons why the results from this study differed from those in the study of Tracy and Enoka (2002). Possibly the most important difference involved the method of measuring steadiness. Tracy and Enoka (2002) measured angular displacement standard deviations rather than acceleration standard deviations. They admit that this measure may be less sensitive than acceleration based measures and so this may partially explain their results.

Another major difference was the level to which target force was normalised. In contrast to our absolute fixed loads of 1kg and 5kg, Tracy and Enoka (2002) used normalised loads that were 5, 10 and 50% of the 1RM concentric load. In this sense their methodology could be regarded as superior to that in this study, as explained earlier. Since both groups had the same absolute target levels in the present study, it was not necessary to normalise SD to the absolute target force. Tracy and Enoka (2002) also did not normalise their SD measures, but this was despite each group having different absolute force levels. They found that SDs actually fell with increasing force levels, which indicates that the stronger, younger, subjects will have shown less unsteadiness than otherwise. Hence this does not appear to threaten their conclusions. However, the present study and that of Schiffman and Luchies (2001) have found that SDs during anisometric contractions do rise with increasing force level, and so their results may still be in doubt.

Schiffman and Luchies' (2001) conflicting findings of no age effects may also be attributable to methodology. As in Tracy and Enoka (2002), steadiness was not normalised, although relative target forces were used. The target forces of 20 and 60% MVC were probably relatively higher than those in the present study, which may also explain the differing results. As in Hortobagyi et al. (2001), subjects had to strive for a steady force rather than a steady velocity, which provides a different challenge. Schiffman and Luchies (2001) used isometric and concentric methodologies that were similar to each other, permitting a direct comparison of steadiness across the two contraction types. It was found that concentric contractions led to more force variability. It was concluded that concentric contractions may therefore be a more sensitive measure for detecting age effects. This may partially explain the results from the present study.

In summary, the findings from this study partially support the findings of Hortobagyi et al. (2001). The cautious nature of that statement is necessary in the light of the differing methodologies. Taken together, both studies suggest that the elderly have worse anisometric steadiness than the young when target forces are not normalised. Further studies are necessary to establish if the same conclusion would apply with normalised target forces. Conflicting knee extensor studies (Tracy and Enoka 2002, Schiffman and Luchies 2001) do not seem to threaten these conclusions, as their methodologies were very different.

5.4.2.ii Effects of load and contraction type on age differences

The lack of group differences for the lower load of 1kg in the presence of differences at 5kg concurs with the findings of Schiffman and Luchies (2001), who noted that higher load concentric contractions lead to a greater effect size and may thus be a more sensitive indicator of age differences (Schiffman and Luchies 2001). The worse test-retest reliability as expressed by LSD at the 1kg compared to the 5kg loads (Appendix 6) also suggests that measurement errors may be partly responsible for the lack of findings at the lower load. However when calculated as CoV, test-retest reliability was equal across loads.

This study is unique in its finding that concentric contractions showed a greater age difference in steadiness than eccentric contractions. In contrast, Hortobagyi et al. (2001) showed that eccentric contractions are more affected by age, and similar findings are reported in the hand and elbow muscles (Graves et al. 2000, Laidlaw et al. 2000, Burnett et al. 2000).

There is some theoretical support for the opposing findings (Hortobagyi et al. 2001, Graves et al. 2000, Laidlaw et al. 2000, Burnett et al. 2000). The lower activation levels associated with eccentric contractions (Laidlaw et al. 2000, Christou et al. 2003a, Enoka et al. 2003) may lead to lower discharge rates (Graves et al. 2000). Lower discharge rates in old age (Nelson et al. 1984) might mean that a further reduction in rates during eccentric contractions would place motor units on a steeper part of the force frequency curve, where a certain level of firing rate variability would lead to larger variations in force (Graves et al. 2000). Whilst young people might experience similar decreases in firing rate during eccentric contractions, they might be shifted to a less steep portion of the curve due to their greater initial firing rate.

It is important to note, however, that the influence of firing variability on large muscle steadiness has been questioned (Tracy and Enoka 2002, Graves et al. 2000) and so this mechanism may not necessarily apply to the knee extensors. Thus the findings of greater steadiness age differences in concentric contractions are not excluded, although it is difficult to explain this phenomenon and further studies are required to confirm or refute these results.

5.4.3 Functional steadiness

5.4.3.i Standard deviations of acceleration

This is the first study that has assessed steadiness during functional tasks. Two of the four functional tasks – standing from a chair and stepping down - were associated with worse steadiness in older subjects. Hence these tasks might be those that would be most compromised by age in terms of control. Furthermore, the averaged measurements suggest that the very consistent trend towards general greater functional unsteadiness in older subjects was a real effect.

The less steady leg was consistently more affected by age, which relates to the greater steadiness asymmetry observed in the older subjects. This greater asymmetry of steadiness with age corresponds to the greater asymmetry in concentric strength with age (Chapter 3). Caution is required before speculating on the implications of this result, as the finding was only observed individually during stepping down. However, this may imply that force control degeneration may occur focally rather than diffusely.

Strength did not correlate with the variables showing a group difference, excluding the possibility that the steadiness differences were a result of strength differences. The speed of movement was not controlled, and hence this was also a possible confounder, as faster velocities have been shown to lead to greater unsteadiness (Christou et al. 2003a). Subjects were allowed to select their preferred speed for pragmatic reasons and because adopting an unnatural speed could itself act as a confounder. However, it is unlikely that the older subjects moved more quickly than the young, given their lower power and strength (Chapter 3), and so it can be assumed that the greater acceleration fluctuations in the older subjects were not the result of differing speeds. What is more likely is that given that the older subjects possibly moved at slower speeds, the differences detected between groups may have been reduced. In turn this suggests that the lack of functional steadiness differences between groups in sitting and standing up could be partially due to slower movement by the older subjects. Whilst it is pragmatic to use preferred speeds, further work should measure movement speeds to test these hypotheses.

Given the differences between young and old for steadiness during standing, it was surprising that steadiness during the similarly concentric stepping up task was also not affected by age. The standing task involved larger knee angles with greater movement through the length/tension relationship and possibly greater recruitment of the gluteal and hamstring muscles. These factors may have contributed to the different findings. The finding that steadiness when stepping down was more adversely affected by age than stepping up appears to support evidence from other studies showing that eccentric contractions demonstrate more of an age effect (Hortobagyi et al. 2001, Graves et al. 2000, Laidlaw et al. 2000) (although this does contradict the anisometric results). Because of this, it is perhaps surprising that steadiness in the eccentric task of sitting did not show age effects. Reliability across the functional measures was uniformly good when expressed as CoV, but the LSD value for sitting was much higher than for the other functional steadiness tests (Appendix 6). Hence this may partly explain the lack of findings for this task.

One methodological limitation should be discussed. Stepping down involved a backward step rather than the more functional forward step. This was performed for safety reasons, as a forward step would not have permitted the subjects to place their hands on to the support pole, and the trailing EMG wires might have led to a trip. Whilst both forward and backward steps involve the quadriceps and gluteal muscles working eccentrically it is possible that subtle biomechanical differences may have led to different results. A side rail on the non-camera side and wireless EMG may have avoided the need to use a backward step, and such alternatives should be considered in further work.

5.4.3.ii Fourier analysis

Standing and stepping down were also the only manoeuvres to show greater acceleration power in older subjects when the acceleration data were separated into spectral bands. The frequency band at which this occurred for the stand and step down manouvres (4-8Hz and 18-32Hz respectively) does not suggest that this age effect was due to differences in the accelerations produced by the gross volitional manoeuvre itself. Given the speed at which most subjects performed the manoeuvres, such gross accelerations would probably have produced variations of accelerations within the 1-4Hz band. The greater power in the 4-8Hz band during standing and the 18-32Hz band during stepping down in older subjects may therefore have directly resulted from the greater fluctuations in angular acceleration observed in older subjects being manifested most strongly at these bands.

This is the first study to compare spectrally analysed acceleration or force fluctuations in young and older subjects in the quadriceps, but other studies have compared them in the FDI (Galganski et al. 1993, Erim et al. 1999, Burnett et al. 2000, Christou et al. 2003a), muscles involved in the pinch grip (Lazarus and Haynes 1997) and in middle finger extensors (Birmingham et al. 1985). Increases in power around 4-8Hz (Erim et al. 1999), 6-10Hz (Galganski et al. 1993) and 2-4 and 10-14Hz (Lazarus and Haynes 1997) were noted in older subjects during isometric contractions, but Burnett et al. (2000) and Birmingham et al. (1985) did not observe age differences in amplitude during anisometric and isometric contractions respectively. In contrast, Christou et al. (2003a) observed that the young had greater power between 0-15 Hz during concentric and eccentric FDI contractions. These conflicting results could result from the similarly conflicting greater SD of acceleration observed in the young group.

Galganski et al. (1993) attributed the greater dominant peak of 6-10Hz in the elderly to the larger motor units in older subjects generating higher unfused tetani at the firing frequency (Galganski et al. 1993). It is possible that such a mechanism may underlie the increased power in the 4-8Hz band during standing in this study, though it is unlikely to relate to the higher frequencies seen in the step down. Possibly these higher frequencies could relate to an increase in rhythmic cortical inputs in this frequency range (McAuley et al. 1997) in older subjects.

It is important to note that whilst finger joint acceleration fluctuations may equate to the associated muscle force fluctuations (Kakuda et al. 1999), this may not be the case in larger joints, because of greater extremity inertia, with acceleration frequencies likely to be lower than the muscle fluctuation frequencies (Kakuda et al. 1999). Hence it is possible that the knee angular acceleration fluctuation frequencies may be somewhat lower than the muscle force fluctuation frequencies.

5.5 Conclusions

Ageing appears to result in a decrease in quadriceps concentric and functional steadiness, but not isometric steadiness. Dynamic and particularly functional methods of evaluating steadiness may therefore be more sensitive at detecting age differences. Also, it may be that ageing has a greater deleterious effect on force control during movement than during static contractions. The greater complexity of a movement compared to a static task makes this likely. The decrease in dynamic and functional steadiness may possibly influence falls and function, and the following chapter will examine this possibility.

6 Association between falls risk and muscle steadiness

6.1 Introduction

6.1.1 Steadiness and falls risk

Galganski et al. (1993) and Christou and Carlton (2001) have suggested that falls risk may relate to steadiness, but this has never been tested. No detailed mechanism has been put forward to explain this supposed link, although it has been suggested that good control of limb trajectory may be important when negotiating obstacles or walking on or off a raised surface (Begg and Sparrow 2000). Maki (1997) also asserted that motor control deficits may lead to foot placement errors, which could lead to falls.

6.1.2 Implications for this study

There have been no studies investigating the possible association between falls and steadiness. The hypothesis is that fallers will have worse isometric, anisometric and functional steadiness.

6.2 Methods

6.2.1 Subjects

The older non-faller and faller groups only (Chapter 2) participated in this part of the study. Older non-fallers will be referred to as "non-fallers" in this chapter.

6.2.2 Steadiness tests

The same tests as in Chapter 5 were used.

6.2.3 Statistical analysis

Data were analysed according to steadier and less steady legs, as described in Chapter 5. The group comparison technique is as described in Chapter 2.

To evaluate the effects of strength (quadriceps isometric strength at 80° flexion) and power on steadiness, separate correlations for the female fallers and female non-fallers were performed. Male groups were not assessed because of low numbers. Analyses were performed in separate sex/group categories to ensure that effects of age or sex did not distort relationships.

6.3 Results

6.3.1 Baseline group characteristics and potential confounding variables.

The groups did not differ in age (P>0.05), activity levels (P>0.05) weight (P>0.05) or age-corrected height squared (P>0.05) (Table 4.1, page 85). Sex was included in the analysis as a cofactor. Sex interacted with variables as shown in Table 5.1 (page 136) (these were the same due to a common analysis). Group comparisons were corrected for these interactions.

6.3.2 Normalised isometric steadiness

When bilateral measurements at all contraction intensities were averaged, the fallers had a significantly higher CoV than the non-fallers (Fallers: 2.0±0.1%, Non-fallers 1.4± 0.1%, P<0.01). When variables were anlsyed individually, fallers had greater force CoV (i.e. were less steady) at 25% MVC in both legs and at 10% MVC in the less steady leg, but not at other contraction conditions (Fig. 6.1).

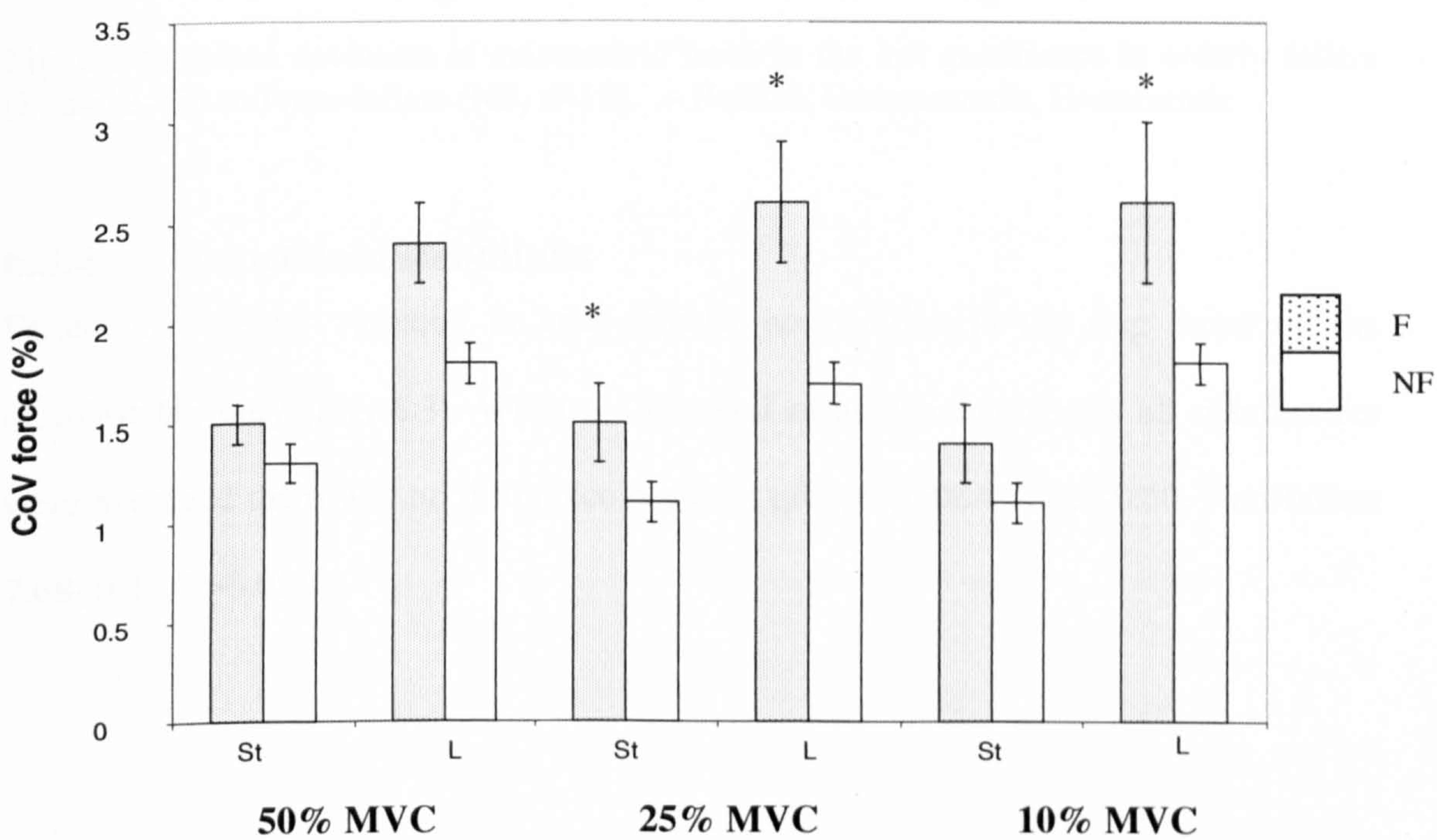


Fig. 6.1 Coefficient of variation of force in the more steady (St) and less steady (L) quadriceps in elderly fallers (F, n=26-28) and non-fallers (NF, n=39-41). * P<0.01

6.3.3 Anisometric steadiness

The fallers had significantly greater acceleration fluctuations during the eccentric 1kg contraction only (P<0.05). When all four measurements were averaged there were no differences between groups (Fallers: 0.031±0.003g, Non-fallers 0.025± 0.002, P>0.05) (Fig. 6.2).

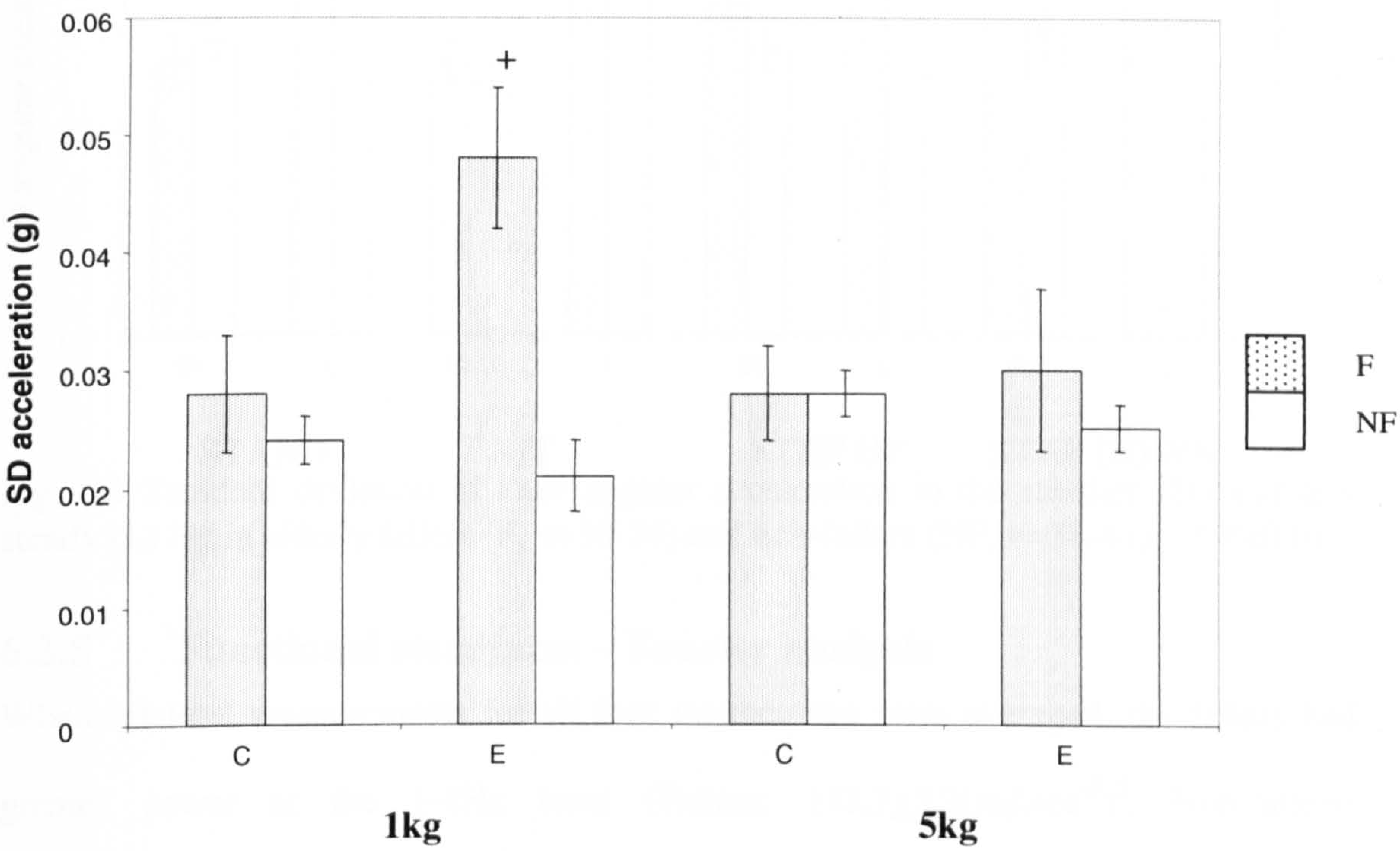


Fig. 6.2 Standard deviation of anisometric force in the left quadriceps in elderly fallers (F, n=12-13) and non-fallers (NF, n=19). + P<0.05. C=concentric, E=eccentric

6.3.4 Functional steadiness

Fallers had greater variation in knee angular acceleration in the step down on the steadiest leg only (Fig. 6.3). When the bilateral measurements across all manoeuvres were averaged there was no difference between groups (fallers: 7.95±0.60, Non-fallers 7.68±0.49, P>0.05].

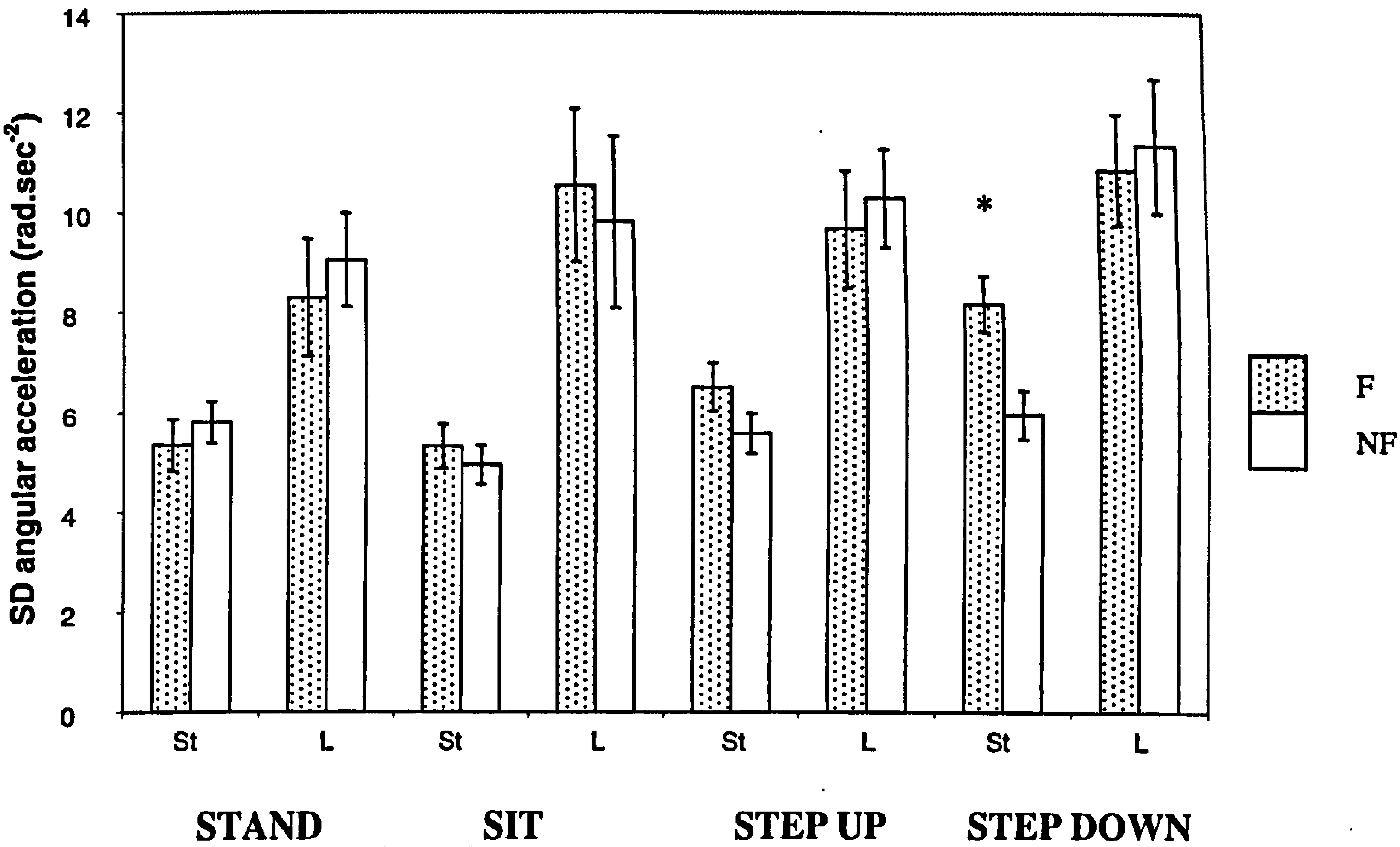


Fig. 6.3 Standard deviation of knee angular acceleration in the steadier (St) and less steady (L) leg in elderly fallers (F, n=30-34) and non-fallers (NF, n=38-41). * $P<0.01$.

6.3.5 Functional steadiness – Fourier analysis

When bilateral measurements for all four manoeuvres were averaged, the fallers had greater power at the 1-4Hz band (Fallers: 113.7±5.2(rad.sec^{-2})^2, Non-fallers: 88.7±4.4(rad.sec^{-2})^2, $P<0.01$) and 4-8 band (Fallers: 102.0±5.8(rad.sec^{-2})^2, Non-fallers: 82.4±4.6(rad.sec^{-2})^2, $P<0.05$) but not at other bands (8-12Hz: Fallers: 89±7, Non-fallers 76±6, $P>0.05$; 12-18Hz: Fallers: 102±13, Non-fallers 112±11, $P>0.05$; 18-32Hz: Fallers: 194±25, Non-fallers 195±18, $P>0.05$; 32-45Hz: Fallers: 61±8, Non-fallers 56±4, $P>0.05$.) When variables were analysed individually, the fallers had significantly higher acceleration power in the step up manouvre in the 1-4, 4-8 and 8-12 Hz bands in the steadier leg, and in the 1-4 and 4-8Hz bands in the less steady leg. For the step down manoeuvre, fallers had significantly higher acceleration power in the 1-4 and 4-8Hz bands in the steadier leg and the 1-4Hz band in the less steady leg. For the sit manoeuvre, fallers had significantly higher acceleration power in the 4-8, 8-12, 12-18 and 18-32Hz bands in the steadier leg (Figs. 6.4-6.7).

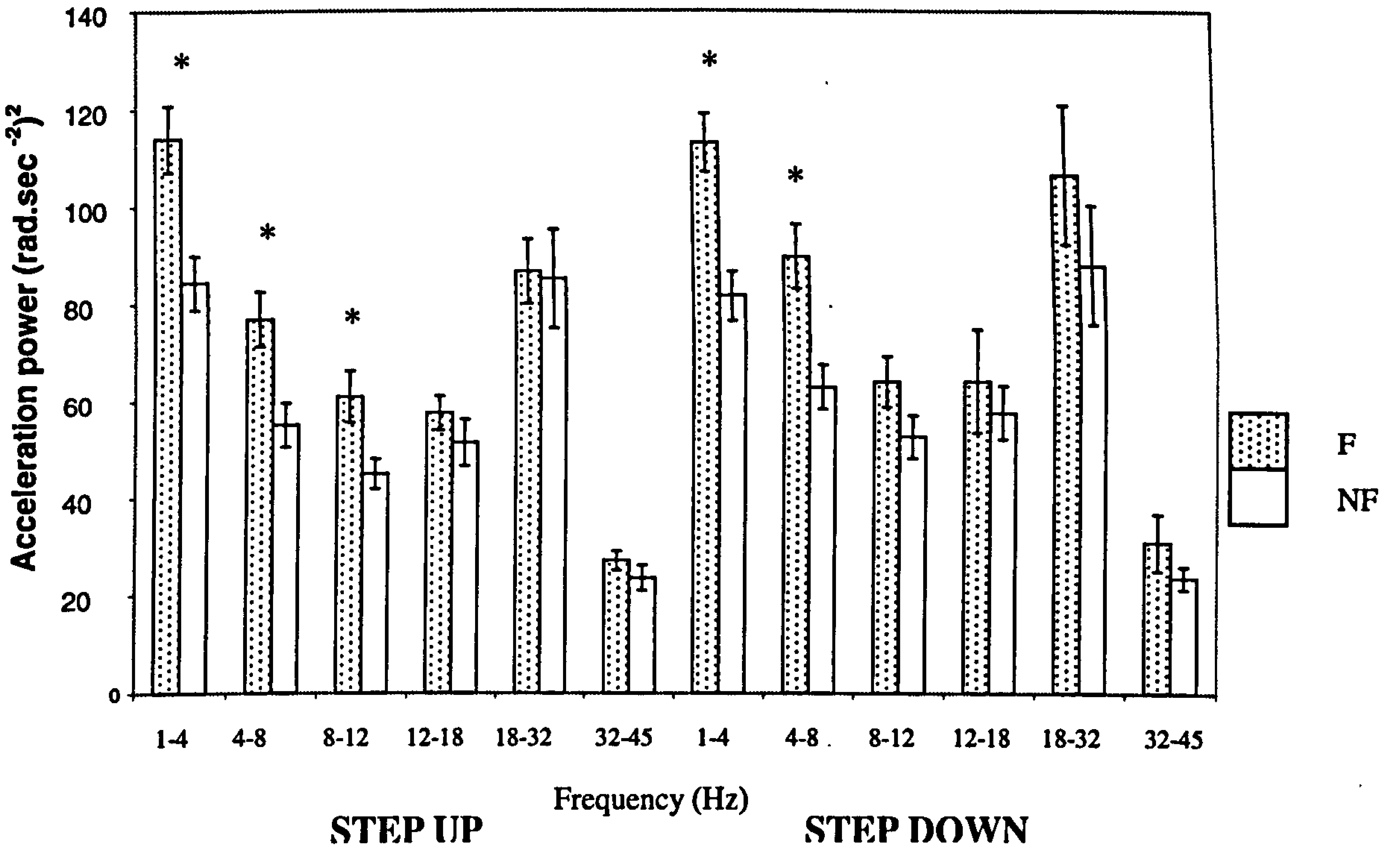


Fig. 6.4 Acceleration power of knee angular acceleration during the step up and step down manoeuvres on the steadier leg at 6 frequency bands in elderly fallers (F, n=30-34) and non-fallers (NF, n=38-41). * P<0.01.

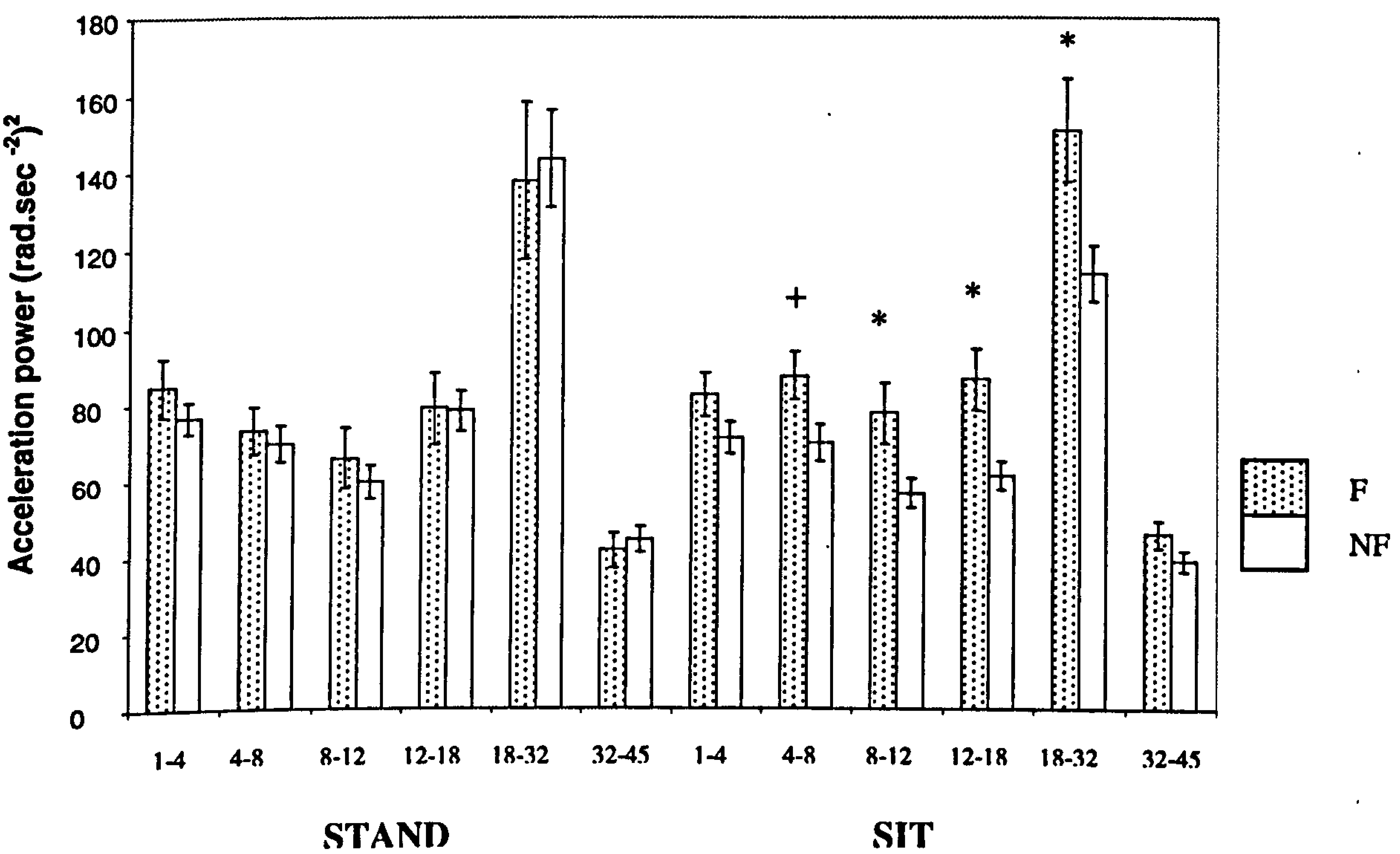


Fig. 6.5 Acceleration power of knee angular acceleration during the sit and stand manoeuvres on the steadier leg at 6 frequency bands in elderly fallers (F, n=30-34) and non-fallers (NF, n=38-41). * P<0.01, +P<0.05.

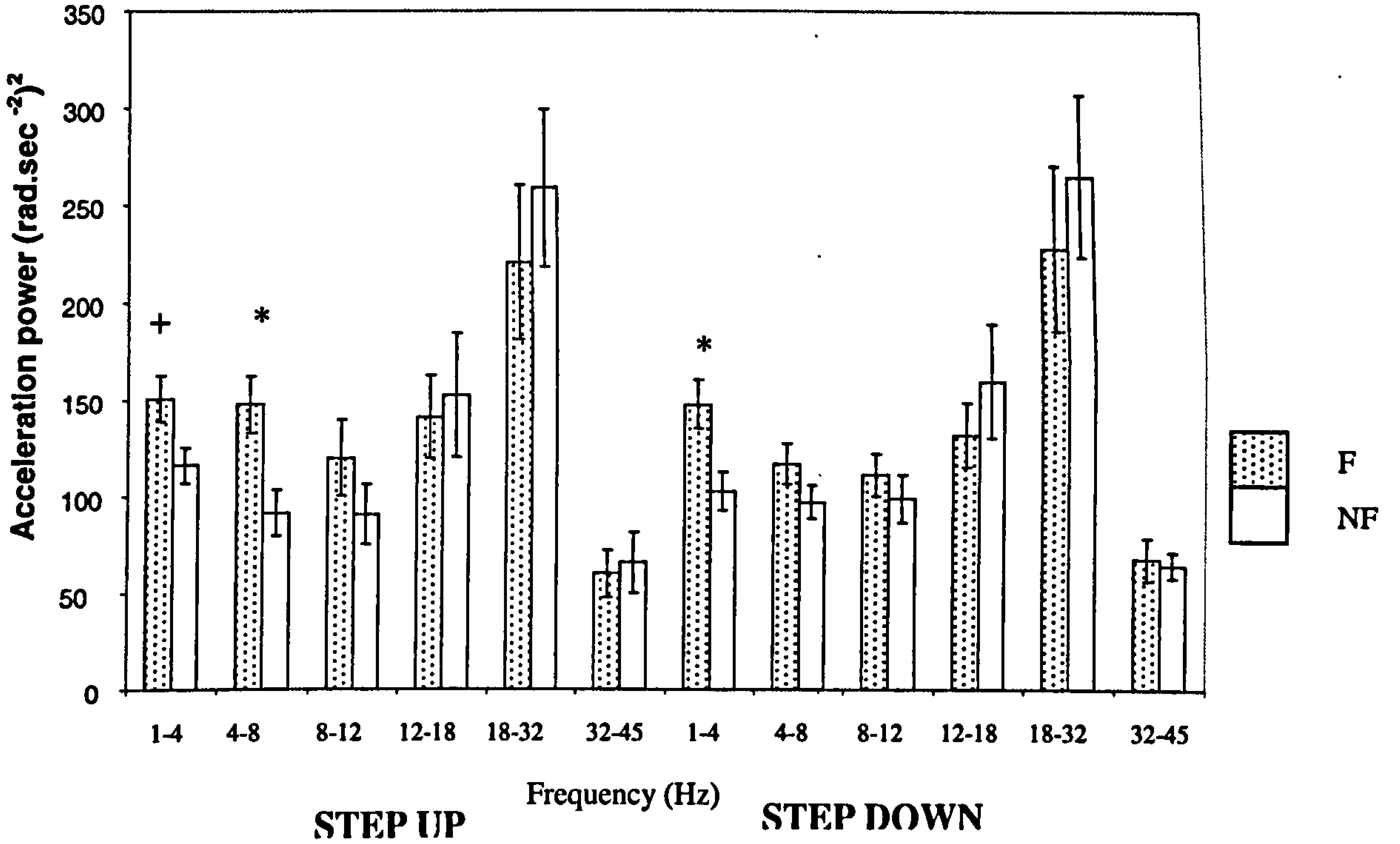


Fig. 6.6 Acceleration power of knee angular acceleration during the step up and step down manoeuvres on the less steady leg at 6 frequency bands in elderly fallers (F, n=30-34) and non-fallers (NF, n=38-41). *P<0.01 + P<0.05.

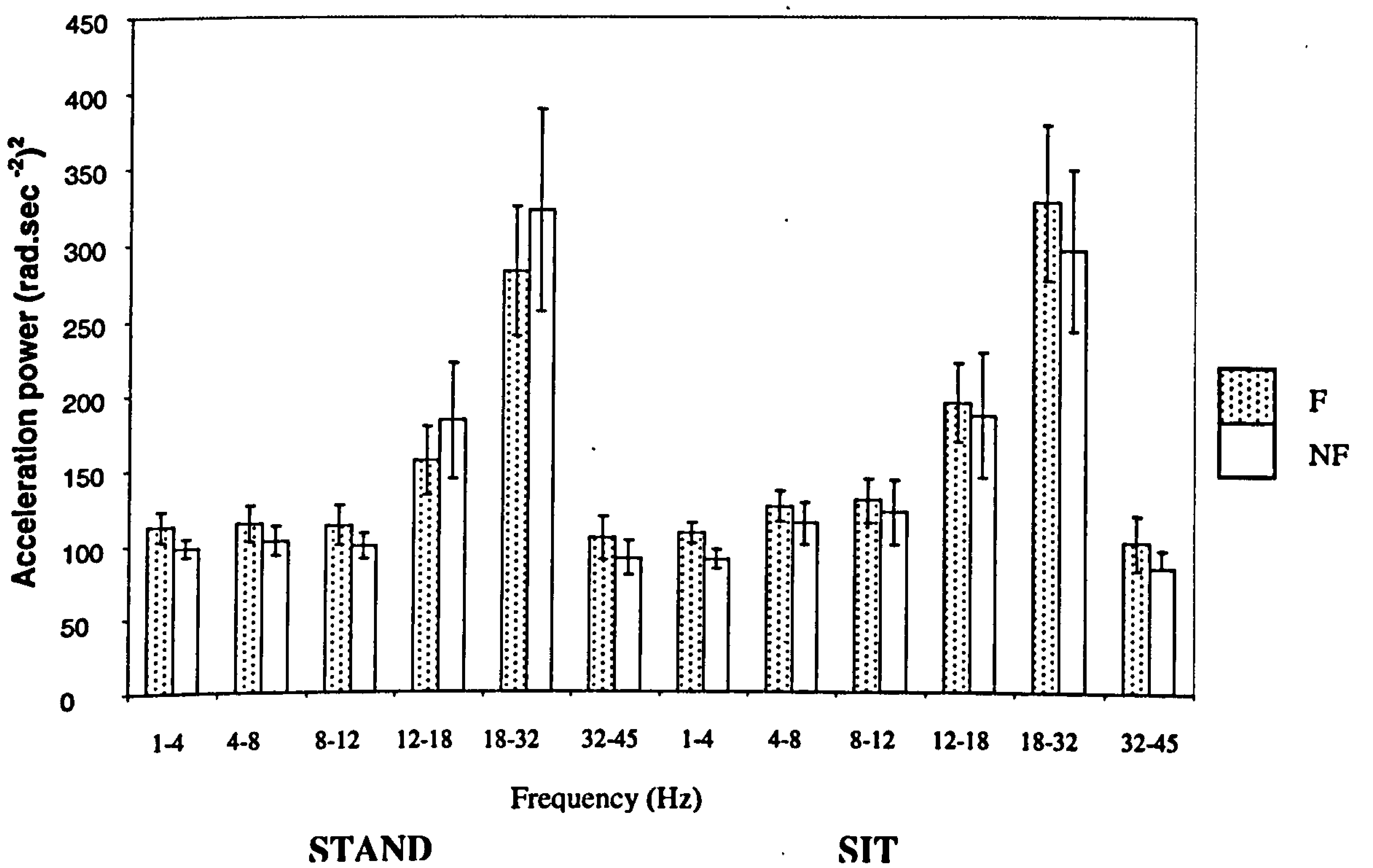


Fig. 6.7 Acceleration power of knee angular acceleration during the sit and stand manoeuvres on the less steady leg at 6 frequency bands in elderly fallers (F, n=30-34) and non-fallers (NF, n=38-41). There were no differences between groups.

6.3.6 Correlation of number of falls and steadiness

When the number of falls in the female fallers group were correlated with steadiness measures, the number of falls correlated significantly with the SD of acceleration during stepping down on the steadiest leg (R=0.039, P=0.04, n=26). The number of falls did not correlate with isometric, anisometric and other functional steadiness measures. The male fallers were too low in number to permit a separate correlation.

6.3.7 Asymmetry of steadiness

The non-fallers had greater asymmetry than the fallers during stepping down only (Fig. 6.8). The Binomial test showed that there was not a significant trend for one group to be numerically greater over the seven variables (P>0.05).

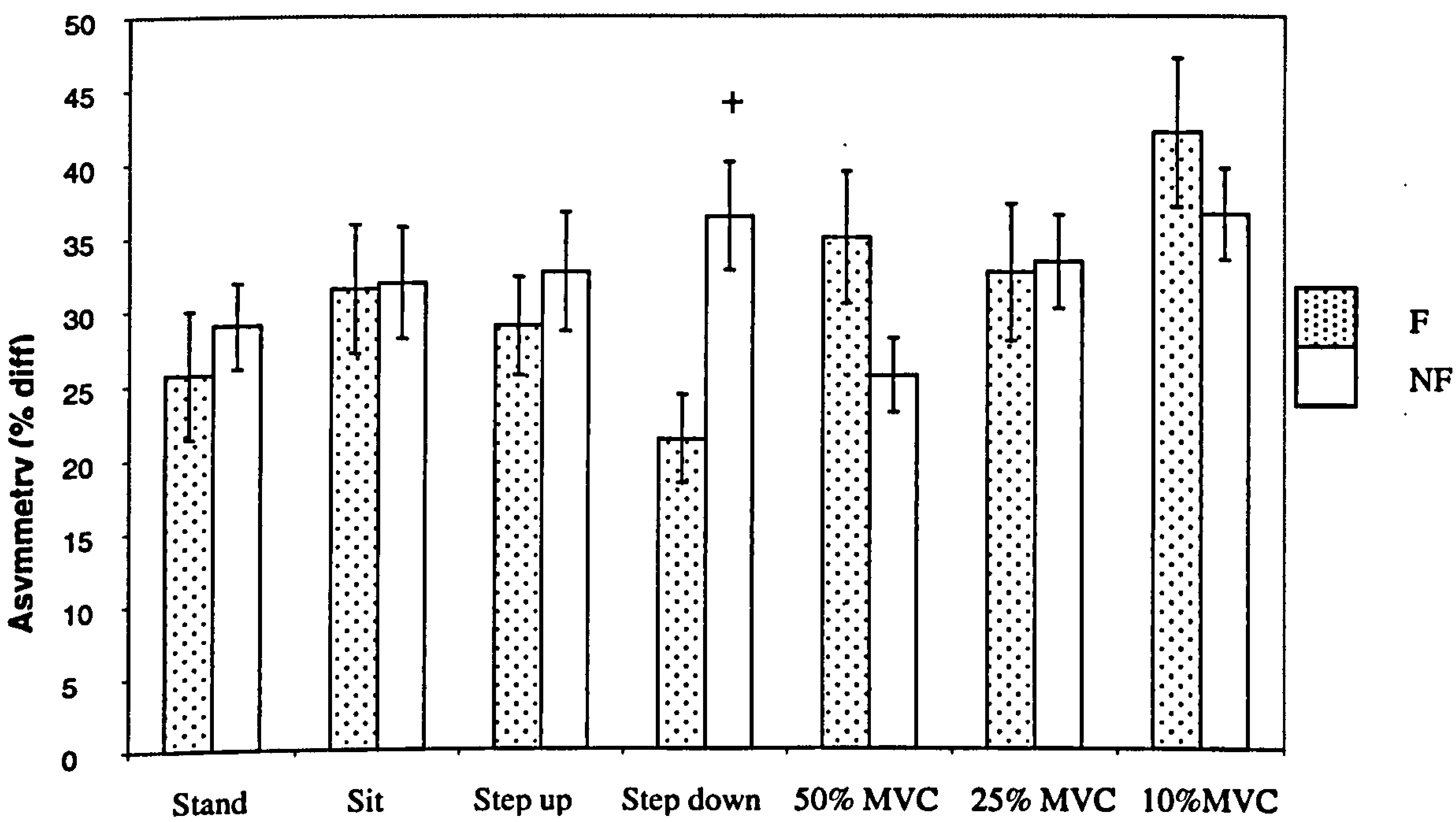


Fig. 6.8 Asymmetry in the functional and isometric steadiness measures in fallers (F, n=25-31) and non-fallers (NF, n=38-43). + P<0.05.

6.3.8 Association between steadiness and strength and power

Isometric peak quadriceps strength correlated negatively with the force CoV at 10% MVC on the less steady leg in the non-fallers (P<0.05) (Table 6.1). Power correlated negatively with the SD of angular acceleration during the step down on the steadiest leg in the non-fallers (P<0.05) (table 6.2). Though it may initially seem inappropriate to correlate low force steadiness with peak strength and power, the aim was to assess any relationship between being steady and being strong/powerful.

Steadiness variable		female fallers		female non-fallers	
		n	R	n	R
Stand	Steady	22	0.13	27	0.14
	Less steady	22	-0.10	27	-0.05
Sit	Steady	22	0.00	27	-0.10
	Less steady	22	-0.10	27	0.01
Step up	Steady	22	0.10	24	0.19
	Less steady	22	0.16	24	0.05
Step down	Steady	22	-0.37	24	-0.17
	Less steady	22	-0.18	24	-0.19
50% MVC	Steady	21	0.35	25	0.17
	Less Steady	21	0.10	25	0.14
25% MVC	Steady	22	-0.01	25	-0.07
	Less Steady	22	0.04	25	-0.28
10% MVC	Steady	21	-0.05	26	-0.39
	Less Steady	21	0.40	26	-0.40+
1kg	Concentric	11	0.29	14	-0.29
	Eccentric	11	0.26	14	-0.17
5kg	Concentric	10	0.05	14	0.50
	Eccentric	10	0.03	14	-0.06

Table 6.1 Association between quadriceps strength and steadiness in female fallers and non-fallers. +P<0.05

Steadiness variable		female fallers		female non-fallers	
		n	R	n	R
Stand	Steady	23	-0.10	29	0.18
	Less steady	24	-0.21	29	-0.03
Sit	Steady	23	-0.11	29	-0.34
	Less steady	24	-0.07	29	-0.07
Step up	Steady	24	-0.13	26	-0.14
	Less steady	23	0.04	26	0.13
Step down	Steady	23	-0.22	26	-0.50+
	Less steady	24	-0.13	26	-0.15
50% MVC	Steady	22	0.13	27	0.09
	Less Steady	22	-0.11	27	0.06
25% MVC	Steady	23	-0.16	27	-0.10
	Less Steady	23	-0.07	27	-0.18
10% MVC	Steady	22	0.10	28	-0.33
	Less Steady	22	0.08	28	-0.15
1kg	Concentric	12	-0.15	15	-0.13
	Eccentric	12	-0.46	15	0.25
5kg	Concentric	11	-0.13	15	0.03
	Eccentric	11	-0.21	15	0.28

Table 6.2 Association between leg extension power and steadiness in female fallers and non-fallers. +P<0.05.

6.4 Discussion

6.4.1 Steadiness and falls

6.4.1.i Isometric and anisometric steadiness

This is the first study to compare steadiness in elderly fallers and non-fallers. The results show that fallers have inferior steadiness to non-fallers at lower intensity isometric contractions, and overall worse isometric steadiness averaged across all isometric tests. Although there was a correlation between better steadiness during the 10% MVC contraction and increased quadriceps strength this is unlikely to explain results as both groups had similar strength at the testing angle.

The difference between groups in steadiness during the 1kg eccentric contraction was marked but isolated, and there was no difference between groups when measures were averaged. This suggests that eccentric force control may be of particular importance in falling. Group differences only in the lighter load are difficult to explain. In isometric contractions, age differences have generally been seen only at lower intensities (Semmler et al. 2000a, Ranganathan et al. 2001a, Laidlaw et al. 2000, Galganski et al. 1993), but this has not been noted during eccentric contractions and the opposite effect has been observed during concentric contractions (Schiffman and Luchies 2001).

In the light of these findings, it was surprising that the number of falls did not correlate with the level of low intensity isometric and eccentric steadiness. It is possible that a threshold effect of steadiness may occur, with the same risk of falling whether slightly or greatly below it.

6.4.1.ii Functional steadiness

The fallers had a higher SD of acceleration in the step down on the steadier leg (i.e. were less steady) but this was the only significant functional steadiness finding. Given the number of variables considered and the P value (<0.05 >0.01) the possibility of a false positive result should not be discounted. However the step down SD also differed between age groups (Chapter 5) and frequency bands in the step down also differed between fallers and non-fallers in acceleration power. In addition, the group differences in this eccentric contraction mirror the eccentric differences during the 1kg contraction. This consistency increases the chance that this may be a real finding. Falls during stepping down stairs are common (Startzell et al. 2000) which also supports this association. Furthermore, the number of falls was found to increase as steadiness decreased during stepping down and standing. This suggests that a threshold effect may not be in operation for this variable.

Nevertheless, the limited functional steadiness findings are surprising in the light of the more generalised isometric findings. Since falls usually occur in a functional context, steadiness measured dynamically should be a more sensitive indicator of falls status. Reliability in young people was uniformly high for the functional tests when measured by CoV, though LSD values for sitting were relatively poor (Appendix 6) suggesting that this could have contributed to the lack of findings for sitting. It is also possible that the validity of the different measures may have varied, and this issue will be discussed in Chapter 8. Any variations in movement speed between groups could also explain results. The fallers were less powerful and weaker and so their possibly slower movement speed may have improved their measured steadiness, as described in Chapter 5.

Spectral analysis showed fallers had greater angular acceleration power than the non-fallers at frequency bands of 1-12Hz during stepping up and down, and between 4-32 Hz during sitting. Overall, when all measures were averaged, the fallers had greater power at 1-4 and 4-8Hz. Spectral analysis of acceleration fluctuations may therefore be a more sensitive method of evaluating steadiness in fallers and non-fallers than merely observing the variations in acceleration amplitude, which probably results from specific group differences at certain frequencies not being detectable when all frequencies are considered at once.

The group differences at 1-4Hz may partly relate to greater accelerations in the low frequency volitional movements in the fallers. However, given that the fallers were less powerful, this is unlikely. Hence the lower frequency differences in acceleration power may be due to the fallers experiencing greater unsteadiness at these frequencies. Such fluctuations at 1-2Hz may result from variations in motor unit firing rate being synchronised across groups of motor units (De Luca et al. 1982) (see Chapter 7). Christou et al. (2004) showed that such low frequency fluctuations increased with age, and this is the first study to suggest an increase in fallers. However, the possibility that frequencies of knee acceleration may be lower than any muscle force fluctuations should be borne in mind (see Chapter 5, p. 163) and thus the group differences in low frequency acceleration fluctuations may actually be due to group differences in muscle fluctuations at a higher frequency.

Asymmetry did differ between groups for the step down manoeuvre, with the fallers having less asymmetry. It is an unexpected finding, as asymmetry in power (Skelton et al. 2002) and strength (Chapter 4) has been noted to be higher in fallers. However, this finding was isolated, with no overall difference across all measures, and further work is required to confirm this observation.

6.4.1.iii Steadiness differences between fallers and non-fallers as an extension of age related changes

Steadiness results were qualitatively very different to those between young and older subjects, where no differences were seen isometrically, and more differences were seen functionally (Chapter 5). The mismatch for functional steadiness is still consistent with the concept that fallers are simply physiologically more aged, as this could merely be explained by the physiological age gap between fallers and non-fallers being less than the chronological age gap between young and older subjects. However, the presence of isometric and eccentric steadiness differences between fallers and non-fallers, in the presence of no age effects on these variables, suggests that these differences between fallers and non-fallers are not simply an extension of age differences. Similarly, symmetry of steadiness was better in younger people but worse in non-fallers than fallers, which leads to the same conclusion. This issue, which was also mentioned in chapter 4, will be discussed further in chapter 10.

6.4.1.iv Mechanisms of the association between steadiness and falls

As argued in Chapter 4, associations between falls status and a variable, in this case steadiness during stepping down and low intensity isometric and eccentric contractions, indicate three possibilities. Firstly, it is possible that the act of falling decreased steadiness. For example, a head injury could indirectly affect steadiness by affecting central nervous system influences on steadiness. In addition, musculoskeletal damage as a result of any injury could potentially affect steadiness directly. However, such scenarios were unlikely in this study as injuries were not serious. As noted in Chapter 4, falling could reduce activity levels and this could affect steadiness, but activity levels did not differ.

Secondly, it is possible that the results may be due to a correlative relationship. For example, it may be that the real factors responsible for falling – such as reduced power or reduced strength – may themselves also directly affect steadiness. However, isometric quadriceps strength correlated negatively with only one steadiness measure that was different between groups - the CoV of isometric force at 10% MVC in the less steady leg in non-fallers. Moreover, this was a weak correlation, with the R^2 value implying that only 16% of any increases in CoV could be explained by decreases in strength. In addition this relationship was only observed in the non-faller group. Similarly, power only correlated with one steadiness measure that showed a group difference – the SD of knee angular acceleration during the step down on the steadiest leg in the non-fallers. The R^2 value implied that only 25% of any increases in SD could be explained by decreases in power. In addition, most other studies have not noted an association between steadiness and quadriceps strength (Tracy and Enoka 2002, Tracy et al. 2001, Hortobagyi et al. 2001, Christou et al. 2003b). There may, however, be other causative factors with which steadiness correlates, such as sensory deficits. As stated in chapter 4, it is possible that variations in sensory or integrative deficits may have influenced falls risk. Moreover, there is some evidence that age–related sensory deficits may impair steadiness (Lazarus and Haynes 1997, Kinoshita and Francis 1996). Studies of deafferented subjects support this (Rothwell 1982, Sanes et al. 1984). Hence steadiness may be a correlate of falling through its relationship to the other potential cause of poor sensory input.

Thus the third possibility, that the reduced steadiness in the faller group was a causative factor in falling, largely depends on the strength of the associations between falls, sensory disturbances, and steadiness as described above. Further work on the strength and nature of these associations, as well as a prospective study, would help to evaluate causal relationships more clearly. It should be stressed that if steadiness is a falls risk factor, it is extremely unlikely to be the sole factor, given the fact that strength and power deficits (Skelton et al. 2002, Chapter 4) are strongly associated with falls but do not correlate with steadiness. Steadiness and strength/power therefore may be independent causes of falls.

No mechanisms for a causal link between steadiness and falling have been closely examined in the literature, but it is possible that poorer control of limb trajectory or velocity, which are functional manifestations of poorer steadiness (Christou et al. 2003a), could lead to timing or placement errors of the limbs either prior to, or when trying to recover from, a fall (Begg and Sparrow 2000). For example, poor control of trajectory could lead to a foot being placed slightly closer to an obstacle than otherwise, initiating a trip. It is also possible that greater lower limb unsteadiness during actions such as stepping could directly initiate postural instability by invoking generalised resonant perturbations in the body as a whole. Further studies are required to investigate this.

6.5 Conclusions

Reduced steadiness during isometric and eccentric contractions, and during stepping down, appear to relate to falling. Reduced eccentric steadiness therefore appears important. If reduced steadiness has a causal influence on the likelihood of falling, then this is a potentially important finding. Any strategy that can improve steadiness may then have the capacity to reduce falls risk to some extent. Chapter 9 will examine the role of exercise in improving steadiness, but it is also possible that pharmacological or other agents might improve steadiness, and further work in that area is warranted. Such work requires information on the causes of greater unsteadiness in older people, and the following chapter will examine some possible causes.

7 Possible neurophysiological mechanisms of force fluctuations

7.1 Introduction

Any increased force fluctuations in older people may partly or wholly result from an exacerbation of the mechanisms underlying force fluctuations in the young. In the following sections, evidence of mechanisms for force fluctuations in the young and the means by which age could amplify them will be presented. A description of evidence concerning specific mechanisms for unsteadiness that may be unique to older people will follow.

7.1.1 Decreased steadiness with age as an exacerbation of the mechanisms underlying unsteadiness in the young

7.1.1.i Individual motor units.

Increase in motor unit size

It has been suggested that fluctuations in force output in the young are due to the unfused twitches of motor units that have just been recruited, with fluctuations corresponding to the firing frequency of the motor units (Allum et al. 1978, Christakos et al. 1982). Age related motor unit re-organisation, with its increase in motor unit size, may increase the amplitude of the unfused tetani (Galganski et al. 1993, Laidlaw et al. 2000, Enoka et al. 2003) through summation of a greater number of synchronous fluctuations in the larger number of muscle fibres comprising the unit (Christakos et al. 1982). Motor unit re-organisation may also exert an effect through increasing the step-wise changes in force when recruiting or de-recruiting other motor units (Galganski et al. 1993, Darling et al. 1989).

However, muscle training has been observed to decrease steadiness in the elderly without affecting the size of motor units (Keen et al, 1994) so this mechanism cannot be the only explanation for increases in unsteadiness with age. In addition, Wessberg and Kakuda (1999) have shown that the motor unit firing frequency differs from the frequency of force fluctuations commonly seen.

Variability of action potential discharge

Variability of action potential discharge may lead to force fluctuations in a muscle through the force output from the motor unit being modulated by the varying frequency of action potentials (Laidlaw et al. 2000, Enoka et al. 2003, Taylor et al. 2003).

Nelson et al. (1984) noted that variability of action potential firing increased with age during submaximal contractions, but data on force variability was not collected. Erim et al. (1999) reported that older subjects had greater variability in discharge rate in the FDI at lower forces but did not observe differences in isometric steadiness. The study by Laidlaw et al (2000) was the first to show that such age-related increases in variability were strongly correlated with decreased steadiness at target force contractions $\leq$ 5% MVC. Kornatz et al. (2002) also noted a correlation between firing variability and steadiness of isometric, concentric and eccentric contractions. More recently, Moritz et al. (2005) observed a close association between the decrease in the CoV of force fluctuations and the fall in the CoV of interspike intervals as FDI motor unit force increased, but no correlation analysis was performed. In contrast, despite noting greater unsteadiness in the elderly, Galganski et al. (1993) and Vaillancourt and Newell (2003) did not find age differences in action potential discharge variability, and Semmler et al. (2000a) reported that the young had greater discharge rate variability.

It has been suggested that increased variability of action potential discharge with age may result from less efficient and reliable synaptic transmission from the motor cortex to the muscle (Laidlaw et al. 2000). In addition, lower motor unit discharge rates have been noted in older people (Nelson et al. 1984) and these are associated with greater rate variability (Person and Kudina 1972).

Motor unit number

The loss of motor units with age (Tomlinson and Irving 1977) may also influence force fluctuations. In a simulation study, de C. Hamilton et al. (2004) showed that fewer motor units may lead to greater CoV of isometric force. This may arise because fluctuations in individual units are likely to have less influential effects on the muscle as motor unit number increases (Tracy and Enoka 2002).

There have been no studies into the possible association between age-related steadiness and motor unit firing variability in the lower limb. This may partly be due to the belief that in large muscles with numerous motor units, the influence of single motor units diminishes (Tracy and Enoka 2002). For larger muscles, research has concentrated on mechanisms applying to whole populations of motor units.

7.1.1.ii Mechanisms applying to populations of motor units

Synchronisation of motor units from a common input at the spinal level

Anterior horn cells of two or more motor units may be activated by impulses from a branched common descending pathway, with these branches being of almost equal length (Semmler et al. 2000a). Synchronization may only occur between units with a common spinal input if the anterior horn cells have similar excitatory post-synaptic potentials (EPSPs) (Rothwell 1994, Farmer et al. 1993a). Hence by altering the balance of inhibitory or excitatory inputs to these anterior horn cells the degree of synchronization can be varied; for example synchrony is increased in maximal exertions (Datta and Stevens 1990), in slow concentric (Semmler et al. 2000b) and eccentric (Semmler et al. 2000b, Semmler et al. 2002a, Kornatz et al. 2004) movements, and in attention demanding tasks (Schmied et al. 2000). Synchronisation arising from branched common inputs appears to be more sensitively detected in the time domain (by cross correlation analysis) than in the frequency domain (by motor unit coherence) (Farmer et al. 1993a).

Computer (Yao et al. 2000) and mathematical (Taylor et al. 2003) modelling studies suggest that any increase in motor unit synchronization should elevate the amplitude of overall force fluctuations in the whole muscle because synchronous force fluctuations from individual units (presumably arising from single unit mechanisms such as discharge rate variability) should summate. Accordingly, Semmler et al. (2001) noted a significant correlation between synchronisation and steadiness in a young group, Semmler et al. (2000a) observed a significant correlation in a young group, but not in an old group, and Kornatz et al. (2004) noted a weak but significant correlation in a mixed group of old and young. In contrast, Schmied et al. (2000) and Semmler and Nordstromm (1998) did not observe any relationship between synchronisation and steadiness in young and middle aged subjects.

There is some evidence that cortico-spinal neurones may be lost at a greater rate than spinal motor neurones. Henderson et al. (1980) observed a loss of 49% of 'larger' neurones and 38% of 'smaller' neurones from the motor cortex between the ages of 20 and 90 years. However, Gardner (1940) observed a decrease of only ~25% spinal motor neurones by the age of 70 years, whilst Tomlinson and Irving (1977) noted a similar loss of ~25% spinal motor neurones by the age of 90 years. This limited evidence suggests that a neuroplastic divergence could occur with age which would lead to a greater number of motor units sharing common descending inputs (Semmler et al. 2000a). It has also been argued that increased synchrony may occur with age as a compensatory strategy. Although synchrony may not increase force output in a single muscle (Yao et al. 2000, Semmler 2002b) it may enhance co-ordination of synergist activity (Semmler 2002b), increasing force across several muscles and aiding function.

Four studies have attempted to test directly the idea that the elderly have greater levels of motor unit synchronization. No differences in FDI synchrony between ages were seen in low (Semmler et al. 2000a) or high (Kamen and Roy 2000) intensity isometric, concentric (Kornatz et al. 2003, Kornatz et al. 2004) or eccentric (Kornatz et al. (2004) contractions, whilst Kornatz et al. (2003) noted increased synchronisation in the young during eccentric contractions. Although this appears to indicate that motor unit synchronisation cannot explain the greater force fluctuations in the elderly in the FDI, this cannot be assumed in the lower limb muscles, such as the quadriceps.

Correlation of motor units from a common supraspinal origin

Correlation of motor unit discharge in the frequency domain is believed to result from common oscillatory inputs in the motor cortex (Farmer et al. 1993a, Baker and Baker 2003). Through corticospinal pathways, these cortical influences are believed to feed into groups of motor units at a certain frequency, increasing the probability of motor unit firing at the incidence of each impulse (Brown et al. 2000, McAuley and Marsden 2000). Although each motor unit will not necessarily fire on each impulse, and maintains its own mean firing frequency, it is probable that any firing will occur at the incidence of these impulses (McAuley and Marsden 2000). At any moment, this results in a proportion of the group of motor units firing synchronously with each other and the oscillatory input, which results in clear rectified EMG peaks (McAuley et al. 1997, McAuley and Marsden 2000). For each motor unit there will be a frequency spectrum that reflects not only its own independent firing rate but also the imposed oscillation frequency, the latter tending to result in some interspike distances that equate to the oscillatory frequency (McAuley and Marsden 2000). Separate motor units exposed to the same oscillatory influence will therefore show coherence at the oscillatory frequency (McAuley and Marsden 2000).

Three principal bands of motor unit coherence have been observed in the young and elderly at ~ 10 and 20 Hz (Farmer et al. 1993a, Conway et al. 1994, McAuley et al. 1997, Halliday et al. 1999, Semmler et al. 2003) and ~ 40Hz (McAuley et al. 1997). Motor unit coherence appears to be more prominent during isometric than anisometric contractions (Baker and Baker 2003). Evidence for supraspinal oscillatory influences on motor unit correlation are that EEG and EMG frequencies are significantly coherent at peaks of around 10 and 20Hz (Farmer 1998) and that central nervous system lesions can remove motor unit coherence (Farmer et al. 1993b).

There is some evidence that greater levels of motor unit coherence, arising from common supraspinal inputs, may be associated with greater isometric unsteadiness in the hand and fingers in the young. Significant coherence between the frequency content of motor unit firing and muscle tremor have been observed at ~ 10Hz during slow movements (Kakuda et al. 1999, Wessberg and Kakuda 1999) and ~ 10Hz and 20Hz isometrically (Conway et al. 1995, Halliday et al. 1999) suggesting coherence and tremor are linked. Importantly, these frequencies do not relate to the mean firing rates of motor units (Kakuda et al. 1999, Wessberg and Kakuda 1999, Halliday et al. 1999). A fixed time lag between EMG peaks and the tremor peaks further supports a relationship (McAuley et al. 1997, Kakuda et al. 1999, Wessberg and Kakuda 1999). A computer modelling study has also shown that the presumed precursor of coherence - common oscillatory activity at 20Hz - gives rise to force fluctuations very similar to those seen in experimental subjects (Taylor et al. 2003). In contrast, Kornatz et al. (2004) were unable to show any correlation between coherence and FDI steadiness in a group of young and old subjects, but this may relate to the anisometric contractions used.

A mechanism for the possible relationship between coherence and steadiness may be that although the synchronisation of action potentials between any two units may not be strong, as shown by cross-correlation (Farmer et al. 1993a, Semmler at al. 2003, Semmler et al. 2004), amongst the whole population of units in the group there will probably be enough units firing synchronously at any moment to cause summation of individual motor unit force fluctuations (McAuley and Marsden 2000).

There is also some evidence that coherence may increase with age. Semmler et al. (2003) found that elderly subjects had a significantly greater degree of FDI motor unit coherence than young subjects in the 5-9Hz band, and had an additional oscillation at 12-13 Hz that was absent in the young. These frequencies did not relate to mean firing rates. In addition, the elderly subjects had greater isometric FDI unsteadiness. Unfortunately no correlation analysis to rigorously establish an association between coherence and steadiness was undertaken. Coherence at 1-4Hz has also been shown to be higher in the elderly during eccentric (Kornatz et al. 2003) but not concentric contractions of the FDI (Kornatz et al. 2003). No studies have yet investigated motor unit coherence in the lower limb.

The increased coherence in the 5-9 and 12-13Hz bands in the study by Semmler et al. (2003) is supported by the finding that the young have a decreased EMG spectral peak at 10Hz relative to the elderly and that this is correlated with better steadiness (Vaillancourt et al. 2003). Brown et al. (2000) suggested that lower frequency oscillations lead to greater unsteadiness because any motor unit synchrony occurring at this frequency will lead to summation of unfused tetani and thus lead to larger fluctuations. Altogether, these facts suggest that both the degree of coherence between motor units and the nature of their individual spectra combine to influence steadiness. The interaction of the two is possibly as follows: the spectral content of individual units influences the amplitude of unit force fluctuations whilst the degree of coherence influences the degree to which these fluctuations are summated.

Common modulation of frequency variability

There is evidence that groups of motor units may have synchronised variations in firing frequency at a frequency of 1-2 Hz (De Luca et al. 1982). This has been termed "common drive" (De Luca et al. 1982). This will lead to synchronised force peaks at this frequency (De Luca et al. 1982). The origin of this is unknown, but it is not thought to be cortical (Farmer et al. 1993b). Christou et al. (2004) showed that such fluctuations increased with age, and that stress accentuated this response.

Co-activation

Co-activation of antagonists may also be related to force fluctuations. Burnett et al. (2000) proposed that co-activation of antagonists could enforce a compensatory increase in agonist recruitment of larger motor units, which might lead to increased force fluctuations. In support of this, increased agonist EMG during antagonist co-activation has been shown in arm muscles (Darling et al. 1989). Accordingly, any age-related increase in antagonist co-activation may be a reason for increased unsteadiness in the elderly (Burnett et al. 2000). However, findings have been conflicting. No associations between antagonist co-activation and steadiness have been found in the FDI (Burnett et al. 2000) or the flexor pollicis brevis (Danion and Gallea 2004). In contrast, Tracy and Enoka (2002) reported that the elderly had a greater magnitude of antagonist co-activation during both isometric and anisometric KE contractions at all loads, and that they also had greater isometric unsteadiness. Though a correlation analysis was not performed, this does suggest an association. There is therefore a need for further work on the KE.

Interestingly, some authors have suggested that increased antagonist co-activation may actually reduce force fluctuations through a damping action (Darling et al. 1989, Patten and Kamen 2000, Christou 2003a). Seidler-Dobrin et al. (1998) used a simple model incorporating the relationships between force and force variability, as shown by Galganski et al. (1993), to demonstrate that experimentally observed elbow co-activation in older subjects might improve steadiness. However the model was not described sufficiently to permit evaluation of its suitability. Nevertheless, it is possible that co-activation could simultaneously affect two opposing factors influencing steadiness – damping and increased agonist recruitment – and the net effect on steadiness would depend on the biomechanics of the joint. This may explain differences between joints.

Synergist co-activation

Co-activation of synergists or different heads of the same muscle have also been suggested as factors involved in increased unsteadiness with age. Graves et al. (2000) showed that co-activation of the brachiailis muscle in young subjects helped to reduce force fluctuations during elbow extension through its smaller moment arm. Older subjects did not use this synergist, and thus had greater unsteadiness. Laidlaw et al. (2002), however, showed no effect of non uniform activation of different heads of the FDI on steadiness in older subjects.

Alternating bursts of agonist and antagonist activity

Alternating bursts of activity in agonists and antagonists or synergists may have the direct effect of causing fluctuations in a force vector (Burnett et al. 2000, Enoka et al. 2003), and this has been observed in young adults performing slow finger movements (Vallbo and Wessberg 1993). Darling et al. (1989) also suggested that irregular antagonist bursts might contribute to reduced steadiness in older people. However, alternating bursts that could explain greater unsteadiness in older subjects have not been detected in the FDI (Burnett et al. 2000) or the KE (Tracy and Enoka 2002)

Mechanical factors

Stiles and Randall (1967) showed that positional fluctuations in the finger at rest were consistent with those resulting from an underdamped second order system. This is a system that responds like a spring to a perturbation, with oscillations around the original position slowly decaying over several cycles. Stiles and Randall (1967) proposed that this oscillating system was driven by the asynchronous firing of different motor units, but it is likely that the driving force could equally be any of those previously described, such as motor unit frequency variability or synchrony. Stiles and Randall (1967) also postulated that increased amplitude of fluctuations could result from decreases in the damping of the system. No studies have yet investigated decreased damping of body segments as a possible reason for changing steadiness with age.

Neural feedback loops

It has been suggested that the spinal stretch reflex may also be responsible for finger force fluctuations (Hagbarth and Young 1979) but there is evidence that reflex responses in the fingers do not have adequate strength or timing to explain the usual patterns of fluctuations (Wessberg and Vallbo 1996). In addition, although frequencies of tremor are similar in the upper and lower limbs, distances around the respective reflex loops differ (Halliday et al. 1999). Although stretch reflexes are unlikely to be a direct cause of tremor, the synchronising effect of stretch reflexes may mean that they can have an important modulatory effect (Semmler et al. 2002). Golgi tendon organ reflexes and Renshaw inhibition appear to be unlikely aetiologies of tremor (Wessberg and Vallbo, 1996).

7.1.2 Unique mechanisms for decreased steadiness with age

Alternatively, or additionally, unique mechanisms specific to old age may create additional fluctuations. Burnett et al. (2000) noted differences between the spectra of force fluctuations in the young and elderly, such as a reduction in the peak frequency in the young but an increase in peak frequency in the elderly when loads are increased. Burnett et al. (2000) claimed that such differences suggest that different mechanisms must also be responsible. Known pathological processes such as those operating in Parkinson's disease will not be considered, as they will not be applicable to the healthy sample in this study.

Non –pathological sensory deficits

There is some evidence that sensory deficits may be a natural part of ageing. Reductions in cutaneous receptor density (Quilliam and Ridley 1971) and poorer mechanoelectric transduction in mechano receptors (Schmidt et al. 1990) have been reported with increased age.

Hortobagyi et al. (2001) suggested that impairments in the sensory systems may affect the control of a steady force in the elderly and some investigators have suggested a link between reduced tactile function and reduced fine motor control in old age (Lazarus and Haynes 1997, Kinoshita and Francis 1996). Only one study has attempted to test the hypothesis that impaired sensation and reduced steadiness are linked in the elderly. Ranganathan et al. (2001a) noted a simultaneous reduction in finger steadiness and two-point descrimination in a group of elderly subjects compared to younger subjects. Unfortunately, no correlation analysis was performed and so the strength of any link between sensation and steadiness is unclear.

In contrast, De Serres et al. (2000) reported that greater amounts of visual information actually worsen steadiness in young people. This may be because the subject becomes *over*-reliant on the sensory information and stops using the more predictive control processes that would normally be used (Cole et al. 1998).

Central effects

Ranganathan et al. (2001a) are the only investigators in this field to suggest that age-related effects on the central nervous system, such as non-pathological degeneration of the cerebellum and basal ganglia, may play a role in reducing steadiness. A tendency towards a reduction in the number of central oscillators, manifested by increased lower and decreased higher frequency power in the spectrum of motor unit discharge, is a feature of Parkinson's disease (Vaillancourt et al. 2003). Since the elderly also experience the same shift in power spectra, it has been suggested that an age-related reduction in the number of oscillators via age-effects on the basal ganglia may lead to the increased unsteadiness seen with age (Vaillancourt et al. 2003). This suggestion has not been tested.

7.1.3 Conclusion

None of these theories explaining the age-related decline in steadiness in the elderly appear mutually exclusive and it is likely that more than one mechanism may contribute to any decreased muscle steadiness in the elderly (Taylor et al. 2003).

It is possible that steadiness may not interact with age directly, as perhaps has been implicitly assumed in the literature. For example, it is possible that steadiness may alter with age as a result of mediation by other objective performance factors, such as strength or power. Strength, however, does not appear to be related to steadiness in the knee extensors (Tracy and Enoka 2002, Tracy et al. 2001, Hortobagyi et al. 2001, Christou et al. 2003b) or the FDI (Burnett et al. 2000). Data from the present study supports this, and also suggests that steadiness does not relate to power. Thus it is likely that steadiness is directly affected by age.

7.1.4 Implications for this study

There is a need for further studies on the causes of increased unsteadiness with age in the lower limb, particularly in terms of action potential variability, motor unit synchrony and antagonist co-activation. An investigation into other mechanisms was beyond the scope of this study. The hypotheses were that reduced lower limb steadiness in the elderly may be related to greater:

1. Co-activation of antagonists
2. Motor unit synchrony
3. Variability of action potential discharge.

7.2 Methods

7.2.1 Subjects

For the functional steadiness co-activation study, only subjects from the young and older non-fallers were used (Chapter 2). The older non-fallers will be referred to as "older subjects" in this chapter. For the studies involving needle electromyography, 7 of the young subjects and 8 of the non-falling older subjects were recruited.

7.2.2 Tests

Co-activation measurement during the functional test

During the functional tests of standing/sitting and stepping up/down (see Chapter 5) EMG data from the quadriceps, hamstrings, dorsiflexors and plantarflexors were also collected. Surface EMG electrodes (Biodex Medical Systems, NY, USA) of 1 cm diameter were used to collect EMG data. The data were fed to the EMG aquisition system (Biopac MP100, Biodex Medical Systems, NY, USA) and then to the CODA motion analysis system, where in-built software was used to calculate root mean square (RMS) values.

The electrodes were placed on the quadriceps, hamstrings, tibialis anterior and gastrocnemius muscle bellies. For each muscle, 2 active electrodes were placed along the length of the muscle, whilst an earth electrode was placed on a bony point. Signals from the electrodes were passed to the Biopac MP100 EMG aquisition system, which was set to a gain of 1000 and low and high pass filtered at 10 and 500Hz respectively. Signals were passed to a custom-built analogue-digital converter and then to the CODA system. The sampling rate was 200Hz, which was adequate for RMS data.

The CODA software permitted analysis of the EMG data alongside concurrently collected steadiness data. For all muscle groups, the portion of the EMG data corresponding temporally to the analysed portion of the angular acceleration data during the four functional manoeuvres was root mean squared. EMGs from the same 4 muscle groups were also collected during separate manually resisted maximal isometric contractions of each muscle bilaterally. For the majority of subjects, the examiner was able to apply sufficient manual resistance to fully resist the subjects' maximal force. Dorsiflexion and plantarflexion contractions were obtained at plantargrade, and quadriceps and hamstring maximal contractions were obtained at 45° knee flexion; angles close to the midpoint of the functional procedures. For two of the young male subjects, sufficient manual resistance could not be provided against the quadriceps and hamstrings. They were placed in a strength-testing chair fitted with a padded unyielding plate that could provide very high isometric resistance to the quadriceps and hamstrings at 90 degrees of knee flexion. EMG data from the middle one second period of these maximal contractions were root mean squared.

Levels of activation for each of the muscle groups during each functional procedure were then calculated by dividing each muscle's RMS EMG during the functional procedure by that of each muscle's RMS EMG during the maximal contraction. Co-activation levels were then calculated 1) from the ratios of hamstrings to quadriceps activation (H/Q), and 2) from the ratios of plantarflexors to dorsiflexors activation (PF/DF) and compared between groups. H/Q and PF/DF co-activation levels were correlated with the functional steadiness measures only.

Motor unit synchronisation measurements

Two EMG disposable concentric needle electrodes (TECA N53153, Oxford Instruments Medical, UK) of 0.3mm diameter and 25mm length were used to collect intramuscular EMG data. Each needle electrode has an anode, cathode and ground at its tip. They were connected to a filtering system (Neurolog, UK) with filtering of signals < 30Hz and >

3000Hz with a 50Hz cut-off. A ground lead was applied to the patella. The signal was fed to an A-D converter (1401, CED, UK) and then into a PC, where it was analysed using Spike 2.13 software (Version 4.14, CED, UK).

Only the dominant leg was tested. The subject lay in supine in shorts with the knee resting on a pillow. The skin overlying the rectus femoris belly was cleaned with alcohol and two sterile needle electrodes were inserted into the muscle at different points approximately 2 cm apart. The subject then generated a small isometric quadriceps force, such that a steady stream of action potentials was observed from both motor units for 2 to 5 minutes, as tolerated.

The electrode positions were then moved, to sample other motor units. This was then repeated one or more times.

Spike 2 software (Version 4.14, CED, UK) was used to discriminate single motor unit action potentials from each needle. These were converted into point processes (events). A cross correlation analysis of events from each needle was then performed, with a bin width of 0.001 seconds, each correlelogram containing 200 bins spanning 0.1 seconds either side of the discharge of the reference motor unit (time 0). The cumulative sum technique (Ellaway 1978) was used to determine the position and duration of the central peak. If the peak was not discernable then a peak width of 0.011 seconds, with its centre at time 0, was used to quantify the synchronisation strength. Two measures of synchrony were used:

- The Common Input Strength (CIS), which was the area of the peak above the mean divided by the duration of the correlation. Units are impulses.sec^{-1}

- Kappa (k), which was the area of the peak above the mean divided by the area of the peak below the mean. This is unitless.

Interspike durations were measured manually from each motor unit recording over a 10 second period at the midpoint of the recording. Mean firing frequency and CoV of interspike intervals (CoV_{ISI}) were calculated.

Synchrony measures were correlated with normalised isometric, anisometric and functional steadiness measures that had been obtained in a previous session. Both mean firing frequency and CoV_{ISI} were also correlated with normalised isometric, anisometric and functional steadiness measures.

7.2.3 Statistical analysis

A univariate GLM ANOVA with sex as a cofactor was used, as described in chapter 2, but only the young and non faller groups were represented. Hence no post-hoc tests were used.

Correlations between variables were carried out for the whole sample and young and older groups separately.

7.3 Results

7.3.1 Subject characteristics

Subject characteristics for the co-activation data are as shown in Table 3.1 (Chapter 3, page 45). The other data were collected on a small subset of subjects from the young and older non-faller groups, with characteristics as shown in Table 7.1.

	Young		Older non-fallers		P
	n	Mean (SE)	n	Mean (SE)	
Weight (kg)	7	63.29 (4.02)	8	67.38 (3.14)	>0.05
Age corrected height squared (m^2)	7	2.83 (0.06)	8	3.09 (0.11)	>0.05
Age	7	28.44 (1.99)	8	76.79 (1.35)	<0.01
Male	2		3		
Female	5		5		

Table 7.1 Subject characteristics for the synchrony, frequency and CoV of action potential discharge data.

7.3.2 Co-activation

A Binomial test showed there were no overall trends for the plantarflexor/dorsiflexor (PF/DF) or hamstring/quadriceps (H/Q) activation ratios to differ between groups. Older subjects had greater ratios of activation between the plantarflexors and dorsiflexors (PF/DF) than the young during standing and stepping down on the steadier leg and during sitting down in the less steady leg ($P<0.05$). No other differences were noted. (Figs. 7.1-7.2).

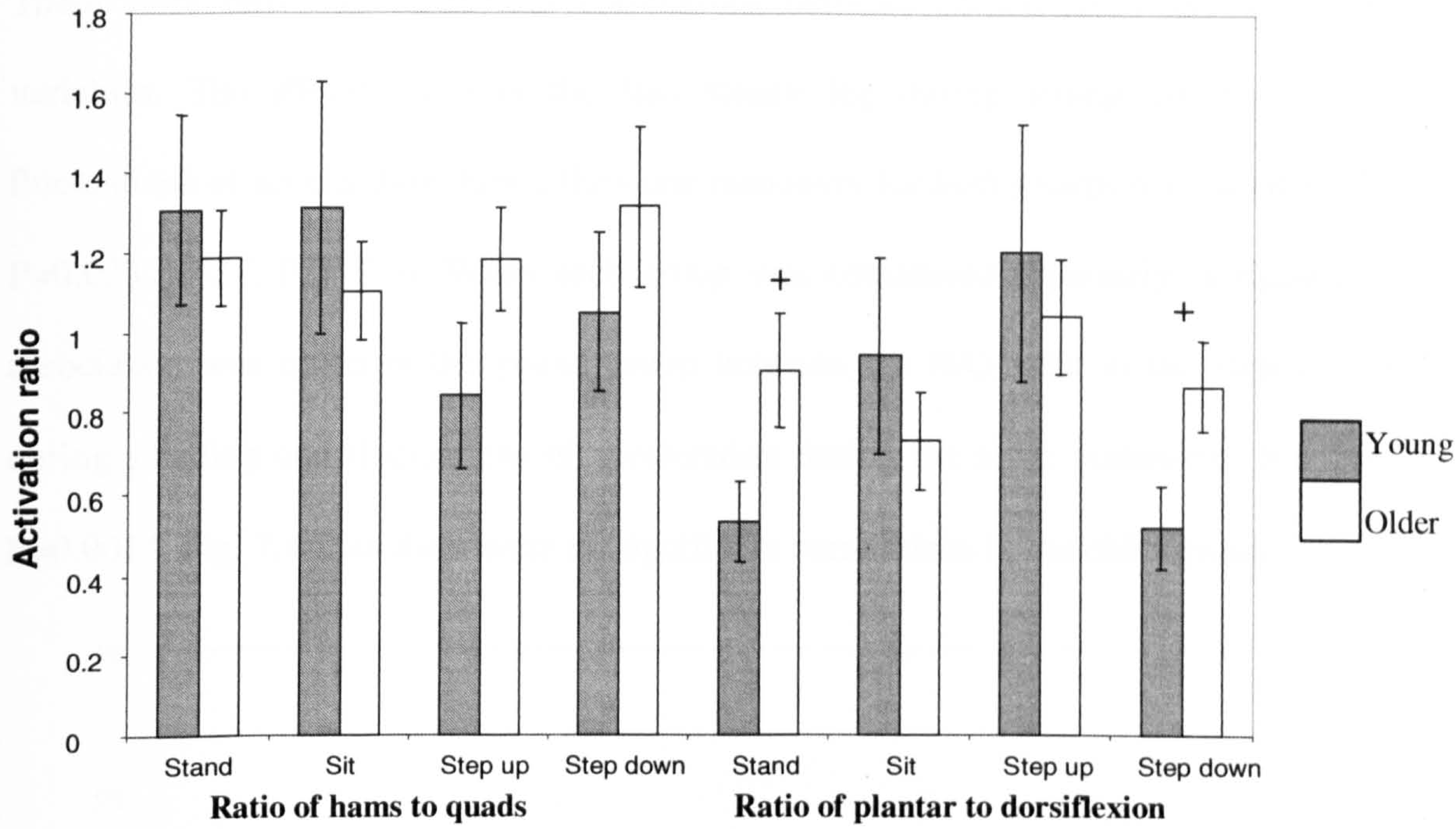


Fig. 7.1 Activation ratios of hamstrings to quadriceps and plantarflexors to dorsiflexors in young (n=19-21) and older (n=19-22) subjects during the stand, sit, step up and step down manoeuvres on the steadier leg. + $P<0.05$.

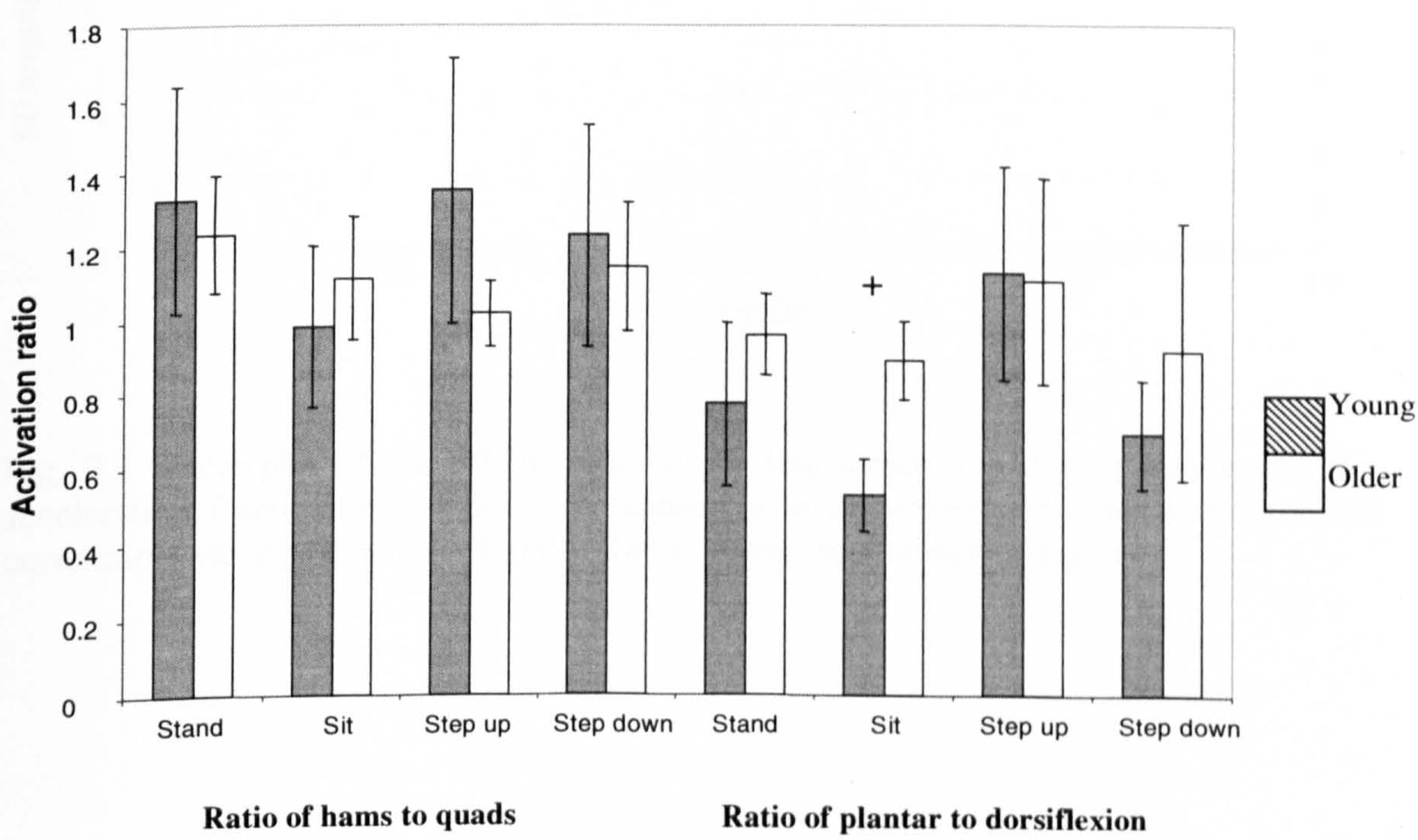


Fig. 7.2 Activation ratios of hamstrings to quadriceps and plantarflexors to dorsiflexors in young (n=19-21) and older (n=19-22) subjects during the stand, sit, step up and step down manoeuvres on the less steady leg. + $P<0.05$.

There were only two significant correlations between co-activation and steadiness variables. The PF/DF ratio in the less steady leg during sitting correlated with fluctuations of acceleration during the same manouvre for both groups overall (R= 0.34, P=0.038, n=37, Fig. 7.3). When each group was considered separately, a significant association was noted in the young group between the H/Q ratio in the steadier leg during standing and fluctuations of acceleration during the same manouvre (R= 0.66, P=0.0014, Fig. 7.4), but there were no significant correlations in the older group.

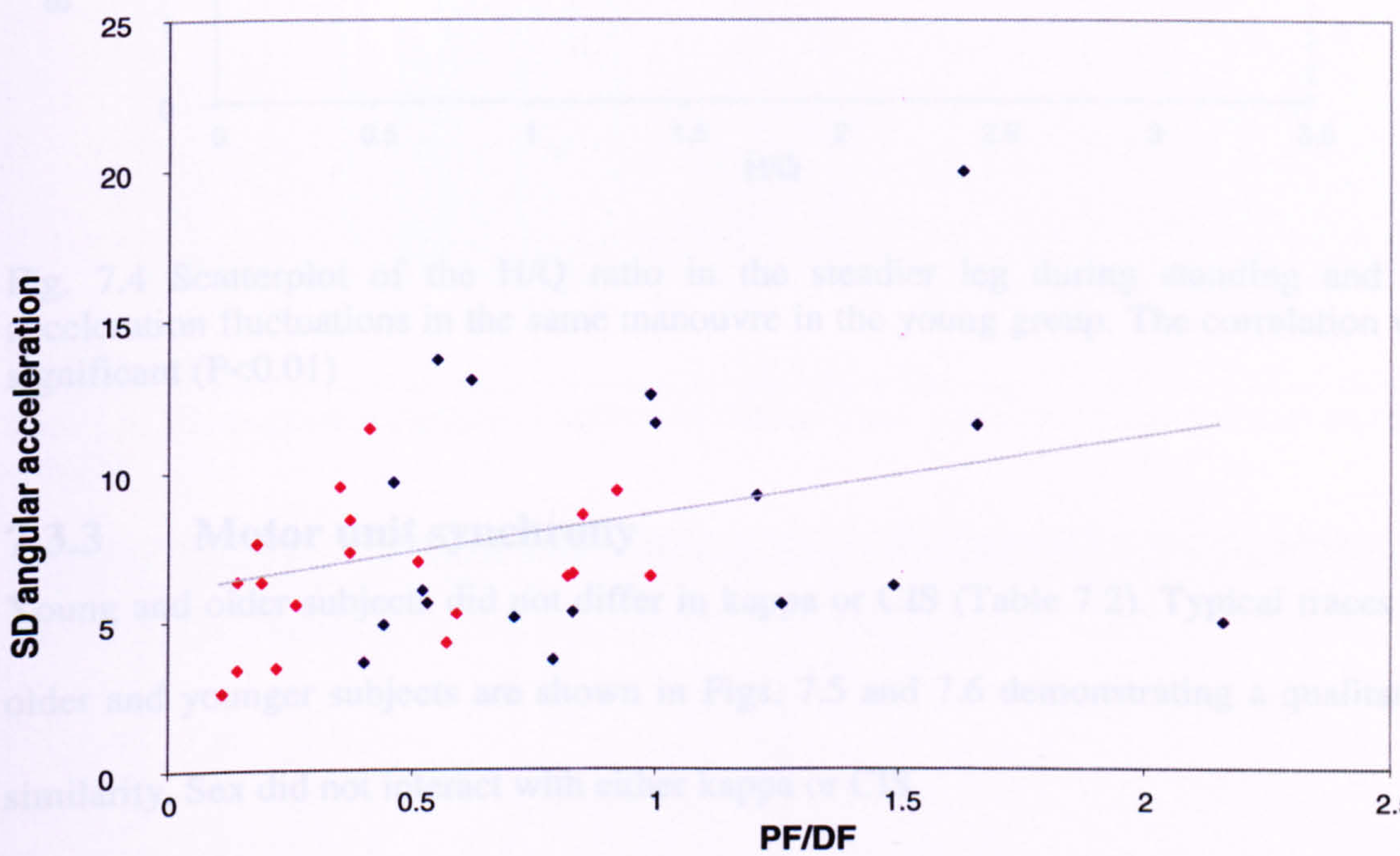


Fig. 7.3 Scatterplot of the PF/DF ratio in the less steady leg during sitting and the acceleration fluctuations in the same manouvre in all young and older subjects. The correlation was significant (P=0.038). Red = young, blue = older subjects

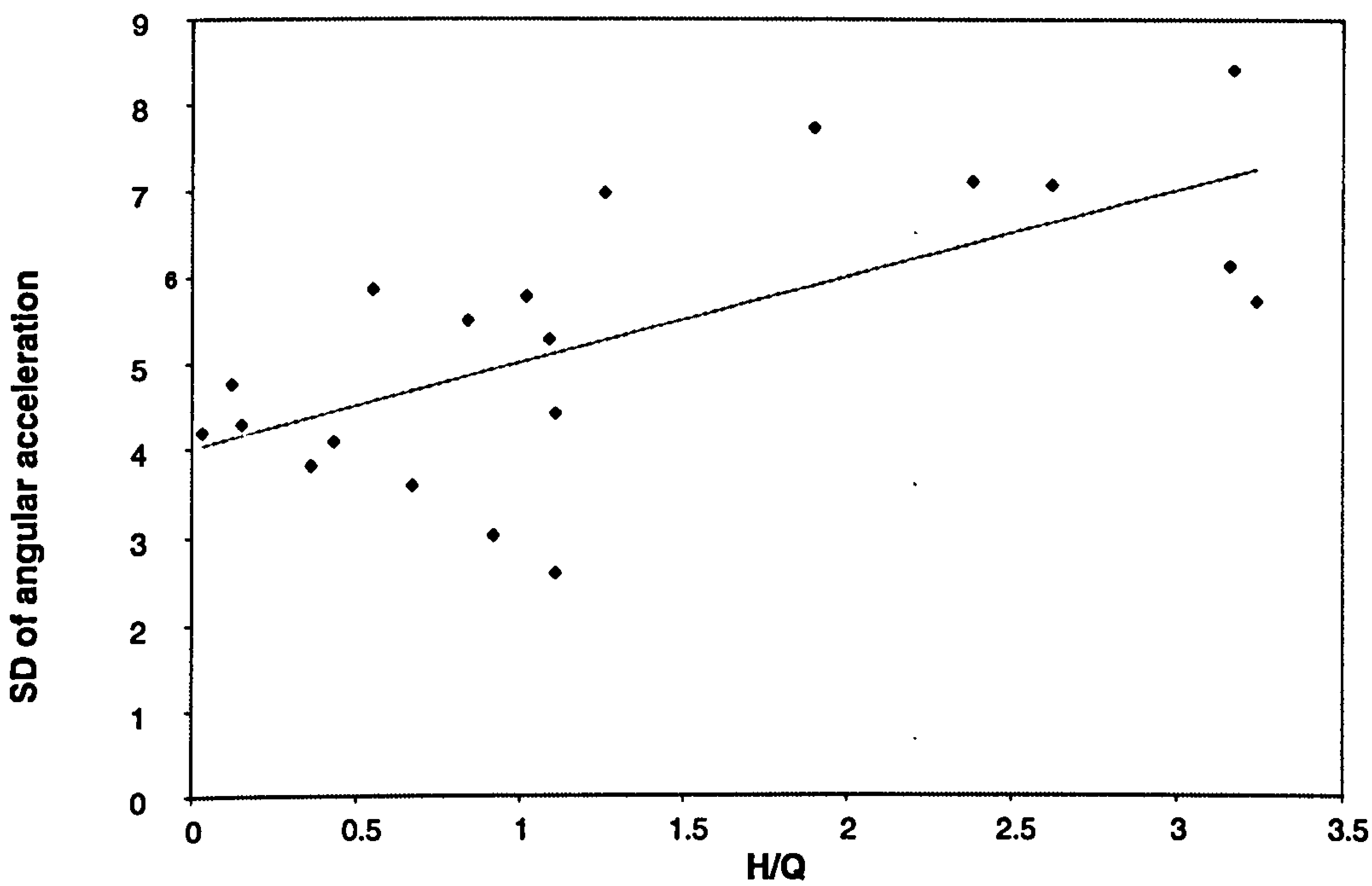


Fig. 7.4 Scatterplot of the H/Q ratio in the steadier leg during standing and the acceleration fluctuations in the same manouvre in the young group. The correlation was significant (P<0.01)

7.3.3 Motor unit synchrony

Young and older subjects did not differ in kappa or CIS (Table 7.2). Typical traces for older and younger subjects are shown in Figs. 7.5 and 7.6 demonstrating a qualitative similarity. Sex did not interact with either kappa or CIS.

		Older		Young
	N	Mean(SE)	N	Mean(SE)
Kappa	8	1.73 (0.45)	7	1.24 (0.13)
CIS (impulses sec^{-1})	8	0.20 (0.05)	7	0.20 (0.09)

Table 7.2. Synchrony in young and older subjects. The groups did not differ in kappa or CIS (P>0.05).

When the whole sample was considered, kappa correlated significantly with acceleration fluctuations during the sit procedure on the less steady leg (R=0.70, P=0.008, n=13), and the concentric 1kg load contraction (R=0.93, P=0.023, n=5). When the groups were considered separately, kappa did not correlate with any steadiness variables.

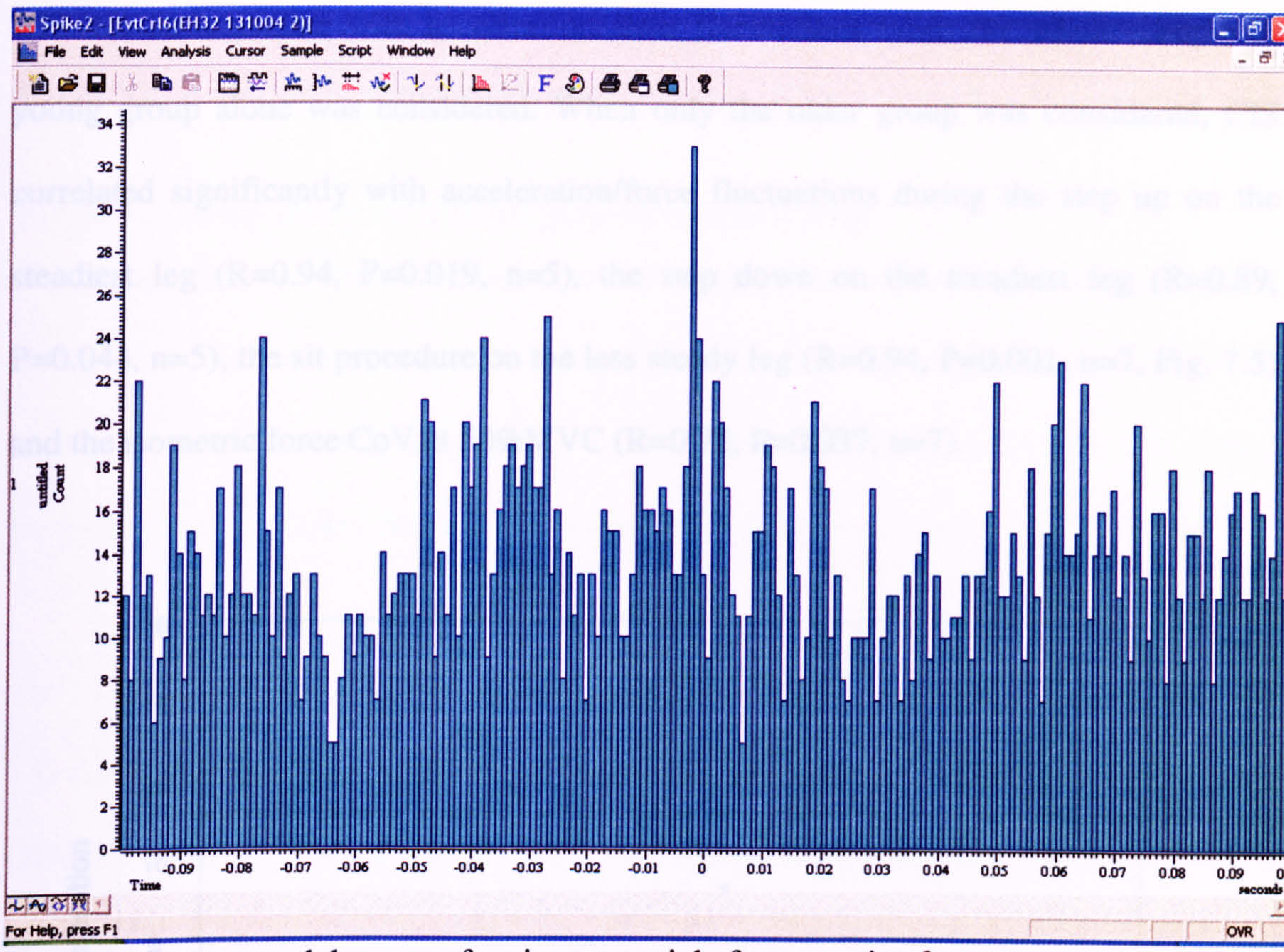


Fig. 7.5. Cross-correlelogram of action potentials from a pair of motor units in an older subject.

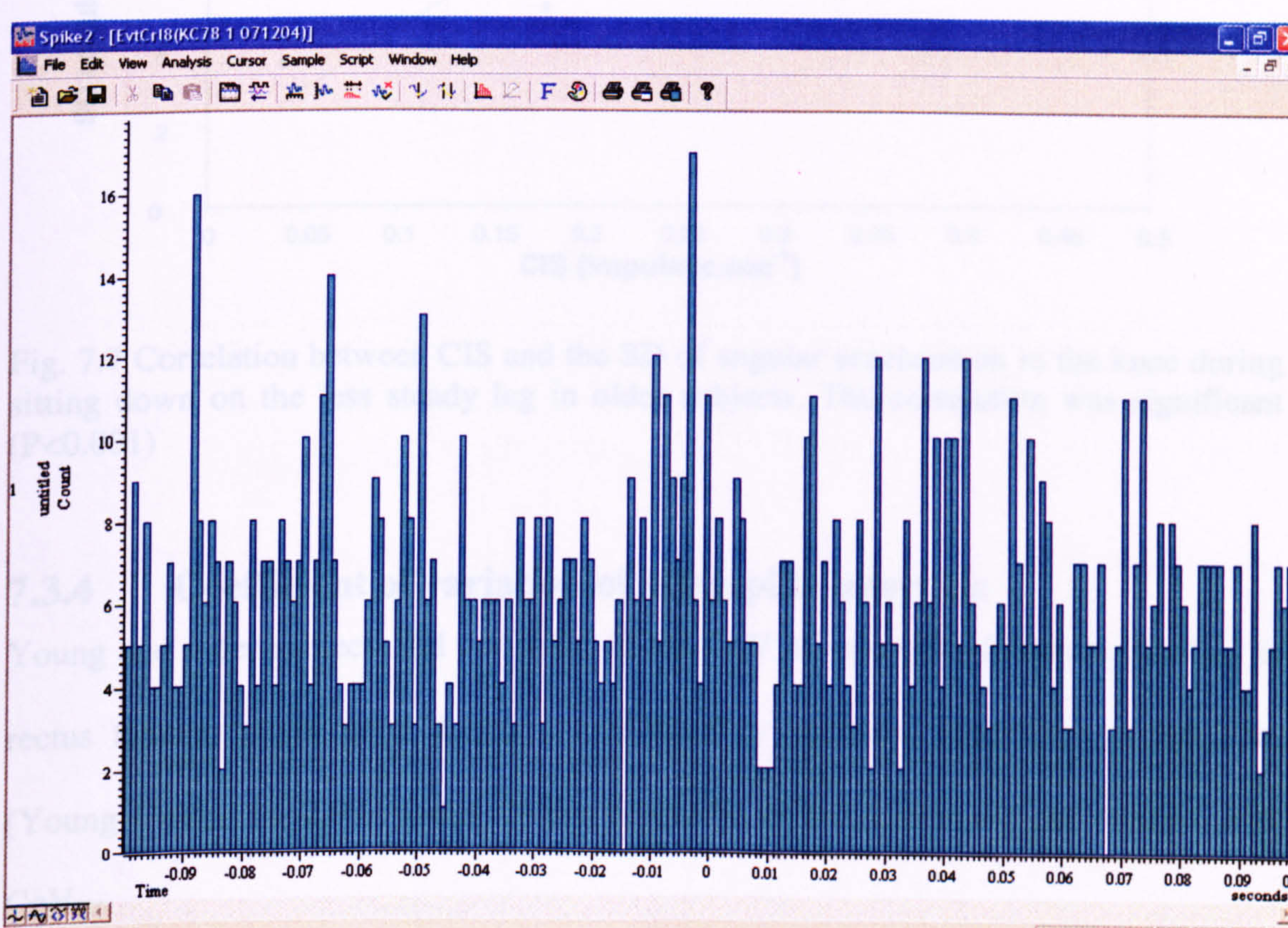


Fig. 7.6. Cross-correlelogram of action potentials from a pair of motor units in a younger subject.

CIS did not correlate with acceleration/force fluctuations when the whole sample or young group alone was considered. When only the older group was considered, CIS correlated significantly with acceleration/force fluctuations during the step up on the steadiest leg ($R=0.94$, $P=0.019$, $n=5$), the step down on the steadiest leg ($R=0.89$, $P=0.044$, $n=5$), the sit procedure on the less steady leg ($R=0.94$, $P=0.001$, $n=7$, Fig. 7.5) and the isometric force CoV at 50%MVC ($R=0.78$, $P=0.037$, $n=7$).

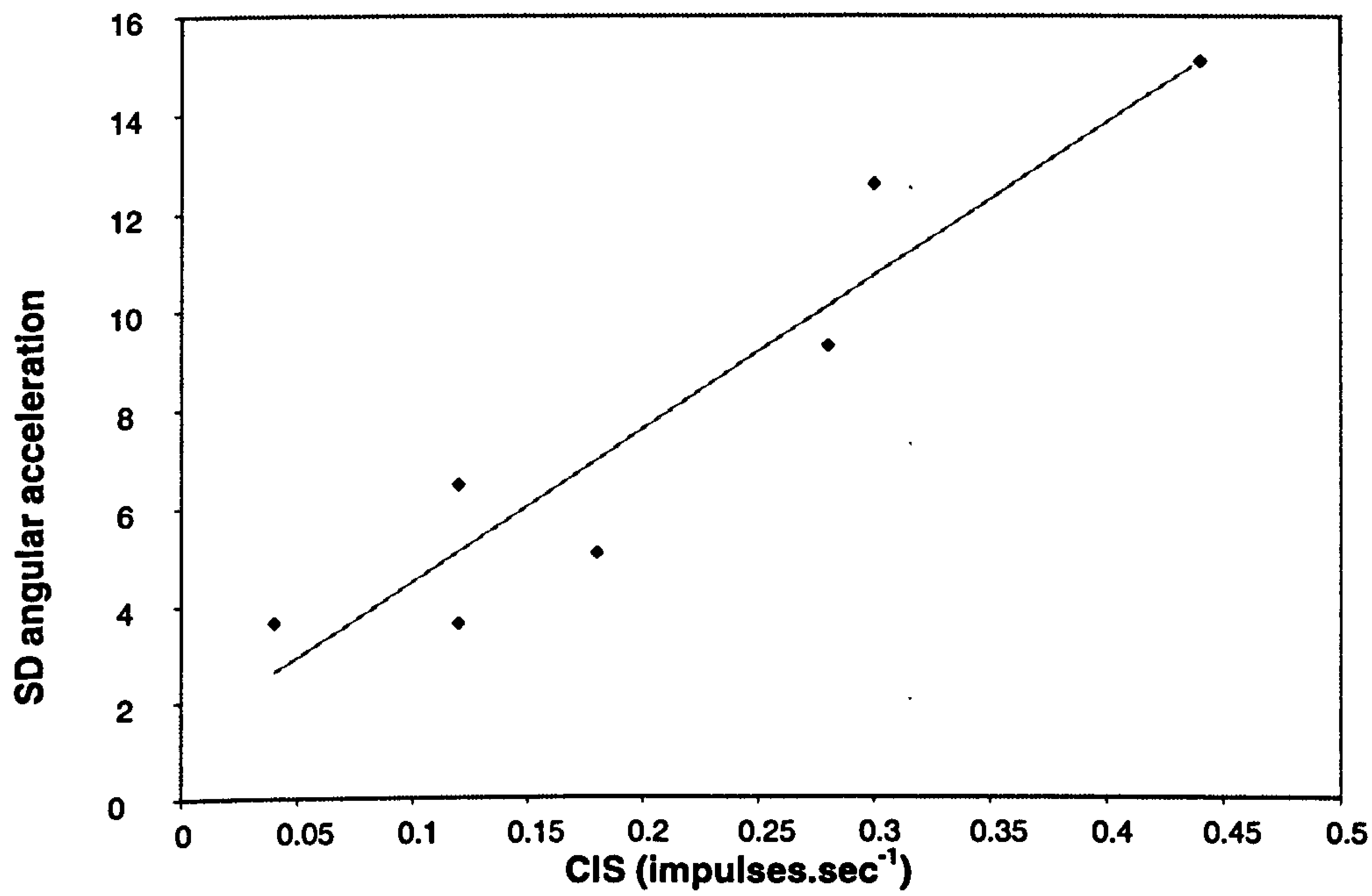


Fig. 7.7 Correlation between CIS and the SD of angular acceleration in the knee during sitting down on the less steady leg in older subjects. The correlation was significant ($P<0.001$)

7.3.4 Coefficient of variation of interspike intervals

Young and older subjects did not differ in the CoV of interspike intervals (CoV_{ISI}) in rectus femoris motor units during a low intensity isometric quadriceps contraction. (Young CoV_{ISI} 14±2.0%, Older CoV_{ISI} 13±1.0%, $P>0.05$). Sex did not interact with CoV_{ISI}.

There were no associations between the CoV_{ISI} and any of the measures of steadiness, for the whole sample or each group considered separately.

7.3.5 Frequency of action potential discharge

The groups did not differ in the frequency of action potential discharge in rectus femoris motor units during a quadriceps contraction at an intensity low enough to allow recording of only one or two motor units (Young frequency 8.76±0.58 Hz, Older frequency 9.35±0.32Hz). Sex did not interact with firing frequency.

The mean frequency of firing did not correlate with the CoV_{ISI} (R=0.34 , P=0.23 , N= 14) for the groups combined. There were also no significant associations for young (R=0.79, P=0.064, n=6) or old (R=-0.18, P=0.70, n=7) considered separately.

7.4 Discussion

7.4.1 Co-activation in the young and elderly

7.4.1.i Knee joint

The lack of differences between young and old in co-activation of the hamstrings and quadriceps during the functional tasks of standing, sitting and stepping up and down, suggests that knee co-activation levels are not important in explaining the differences in steadiness seen between young and old in these tasks (see Chapter 5). The finding that the ratio of hamstring to quadriceps activation (H/Q) is correlated with the steadiness measured during the stand manoeuvre in the young group suggests that force fluctuations in these subjects may be partly due to co-activation, but the lack of such an association in the older subjects underlines the impression that co-activation does not influence elderly steadiness.

In contrast, Tracy and Enoka (2002) reported that the elderly had a greater magnitude of antagonist co-activation during both isometric and anisometric contractions at a number of loads. One possible reason for the difference in results between studies is the differing nature of tasks performed. It may simply be that age may interact differently with co-activation levels in the open kinetic chain movements used by Tracy and Enoka (2002) and the closed kinetic chain movements in this study.

One limitation in the Tracy and Enoka (2002) study was the lack of any correlative analysis between steadiness and co-activation. Although the older group had both worse isometric steadiness and higher co-activation, the lack of a correlation co-efficient means that evidence was not provided that the greater co-activation was linked to worse steadiness. Hence the available evidence suggests that levels of co-activation of the knee flexors and extensors do not influence lower limb steadiness in older people.

7.4.1.ii Ankle co-activation

The fact that older subjects had a higher plantarflexor to dorsiflexor activation ratio (PF/DF) during standing up and stepping down on the steadier leg, and during sitting on the less steady leg is a novel finding. This suggests that age may cause an altered activation pattern in the ankle during these functional movements. Interestingly, greater acceleration fluctuations in the older subjects were also noted during standing up on the steadier leg (see Chapter 5), suggesting an association between age, steadiness and PF/DF. However, there was no significant correlation between acceleration fluctuations during standing on the steadier leg and the PF/DF during the same movement, suggesting no close association.

The significant correlation between PF/DF and acceleration fluctuations in the less steady leg during sitting down for the whole group does suggest that PF/DF may have some influence on steadiness, and because PF/DF increases with age, this may explain age related changes in steadiness. However $R^2 = 0.12$, suggesting any contribution of PF/DF variations to steadiness changes are small, and steadiness differences were not detected between age groups during this task (see Chapter 5). In conclusion, PF/DF probably has little influence on steadiness measured at the knee, although had steadiness been measured at the ankle different results may have been gained.

7.4.1.iii Potential limitations

This study, in common with Tracy and Enoka (2002) and Darling et al. (1989), only used surface electromyography to assess co-activation. Reliance on surface electrodes may have certain drawbacks: synchrony may lead to increases in EMG amplitude (Yao et al. 2000, Keenan et al. 2005) and cancellation of overlapping positive and negative phases can lead to reductions in EMG amplitude (Enoka et al. 2003, Keenan et al.

2005). However, synchrony did not differ between groups, and cancellation of overlapping phases have been shown to have minimal effects when IEMG values are normalised to maximal values (Keenan et al. 2005), as performed in this study.

Some studies have also examined alternating bursts of agonist and antagonist activity as possible causes of greater unsteadiness with age. This was not examined in this study, but this would be a useful focus for future work.

7.4.2 Motor unit synchrony

The lack of difference in motor unit synchrony between groups suggests that this is unlikely to be a significant factor in any age difference in lower limb steadiness. This is the first study to investigate age related changes in lower limb motor unit synchrony, but the results concur with the findings of Semmler et al. (2000a) and Kamen and Roy (2000) in hand muscles.

However, the significant correlations observed between steadiness and both kappa and CIS, in the whole group and the older group alone, suggest that synchrony is an important correlate of steadiness. The positive correlations indicate that as synchrony increases, fluctuations in acceleration increase, and the R^2 values suggest that in older subjects CIS may contribute up to 88% of the variation in steadiness. None of the steadiness variables correlating with synchrony had differed between age groups (Chapter 5) and this concurs with the similarity in synchrony between groups. These results suggest that whilst synchrony may be an important influence on steadiness in older people, it does not explain age differences in steadiness.

It has been suggested that motor unit synchrony could increase with age because of a greater number of motor units sharing common descending input due to a larger age-related loss of corticospinal than spinal motor neurones (Semmler et al. 2000a). This assertion has been made on the basis of only one study assessing corticospinal loss (Henderson et al. 1980) and it is possible that the absence of evidence for greater synchrony in the elderly results from the fact that spinal and corticospinal neurones are not lost at differential rates. Alternatively, or additionally, age–related changes in inhibitory and excitatory inputs to the anterior horn cell may reduce the degree of synchrony expected from any divergence of motor outputs. This adaptation would therefore favour improved force control over the advantage of improved synergy of multiple muscle contractions (Semmler 2002b).

This appears to be the first study to measure synchrony in the quadriceps. The mean values for CIS in this study amongst both groups of around 0.2 impulses.sec^{-1} are more than 3 times smaller than those noted in studies on the FDI (Semmler et al. 2000a), although CIS values of around 0.2 impulses.sec^{-1} have been noted for the FDI in young musicians (Semmler and Nordstrom 1998). Datta et al. (1991) found that the strength of synchrony in the medial gastrocnemius was about one sixth of that in the FDI, whilst in tibialis anterior it was about half of that in the FDI. Datta et al. (1991) concluded that branched stem presynaptic inputs are more active in hand muscles than large limb muscles and the results from this study support this. These conclusions are slightly paradoxical as synchrony might be expected to be less in muscles such as the FDI, which are concerned with fine controlled movement. Synchrony in the FDI may therefore have other advantages which override the disadvantage of decreased fine control. Nevertheless, if the quadriceps does indeed have lower synchrony than the FDI this may partially explain the lack of age difference in quadriceps synchrony; if synchrony is at a low level then any age related changes may be too small to detect.

In contrast to other studies on motor unit synchrony (Semmler et al. 2000a, Semmler and Nordstrom 1998), separate synchrony measures derived from motor unit pairs from single subjects were not regarded as separate members of the sample. Instead the synchrony values derived from single subjects were averaged. Although this may have reduced statistical power in comparison to the other studies, it was felt to be correct as there were large variations in the number of motor unit pairs (1-7) that were sampled among the subjects. Since there was also a large variation in synchrony between subjects within each group this presented a high risk of distortion. Nevertheless, when kappa and CIS were measured using each motor unit pair as a separate member of the sample, overall results were unchanged (kappa P=0.096, CIS P=0.341, young n=17, older n=30).

Limitations of the present study should be mentioned. The only differences in steadiness between age groups were observed during concentric and functional tasks (Chapter 5), but motor unit synchrony was measured during isometric contractions, during which no age differences in normalised steadiness were previously noted. Technically it was difficult to obtain analysable action potential traces during movement, which is why only isometric contractions were used. It is possible that had synchrony been measured during movement an age difference in synchrony conversant with the age differences in dynamic steadiness may have been observed, as Semmler et al. (2000b, 2002) found that synchrony during anisometric contractions was greater than during isometric contractions. In turn, this might explain why only anisometric steadiness differed between age groups. However the significant positive associations between isometrically measured synchrony and the acceleration fluctuations in the anisometric step down task would be unlikely if the isometrically measured synchrony was very different to the synchrony occuring during such an anisometric movement.

Another important limitation is that the levels of synchrony were measured at very different force levels to those occurring during the steadiness measurements, or that might occur during recovery from a fall. There is evidence that synchrony may increase with force levels increasing from 50 to 100%MVC (Kamen and Roy 2000) (although as yet no evidence of the relationship between synchrony and low to moderate forces). Hence it is possible that the measures of synchrony in this study bear little relationship to those seen at more functional force levels. The low force levels used for synchrony measurement were set by practical constraints, as the needle electrodes used tended to detect too many action potential signals than was possible to differentiate when forces were high. Further studies should measure synchrony at higher forces using electrodes capable of more localised action potential detection.

Kappa is a direct measure of the ratio of counted synchronous discharges to those expected by chance (Nordstrom et al. 1992). As such it is a valid measure of synchrony strength. However, it is not independent of firing frequency (Nordstrom et al. 1992), and CIS was developed as an alternative measure that is claimed to be independent of firing frequency (Nordstrom et al. 1992) by being normalised to the duration of data collection. As firing frequencies in this study did not differ, kappa may be regarded as an equally appropriate measurement for this study.

Mention should be made of some very high kappa and CIS values detected between some motor unit pairs in the elderly subjects. These involved central peaks of 1-2 ms width with kappa values up to 32 and CIS values of around 9 impulses.sec^{-1}. An example is shown in Fig. 7.8.

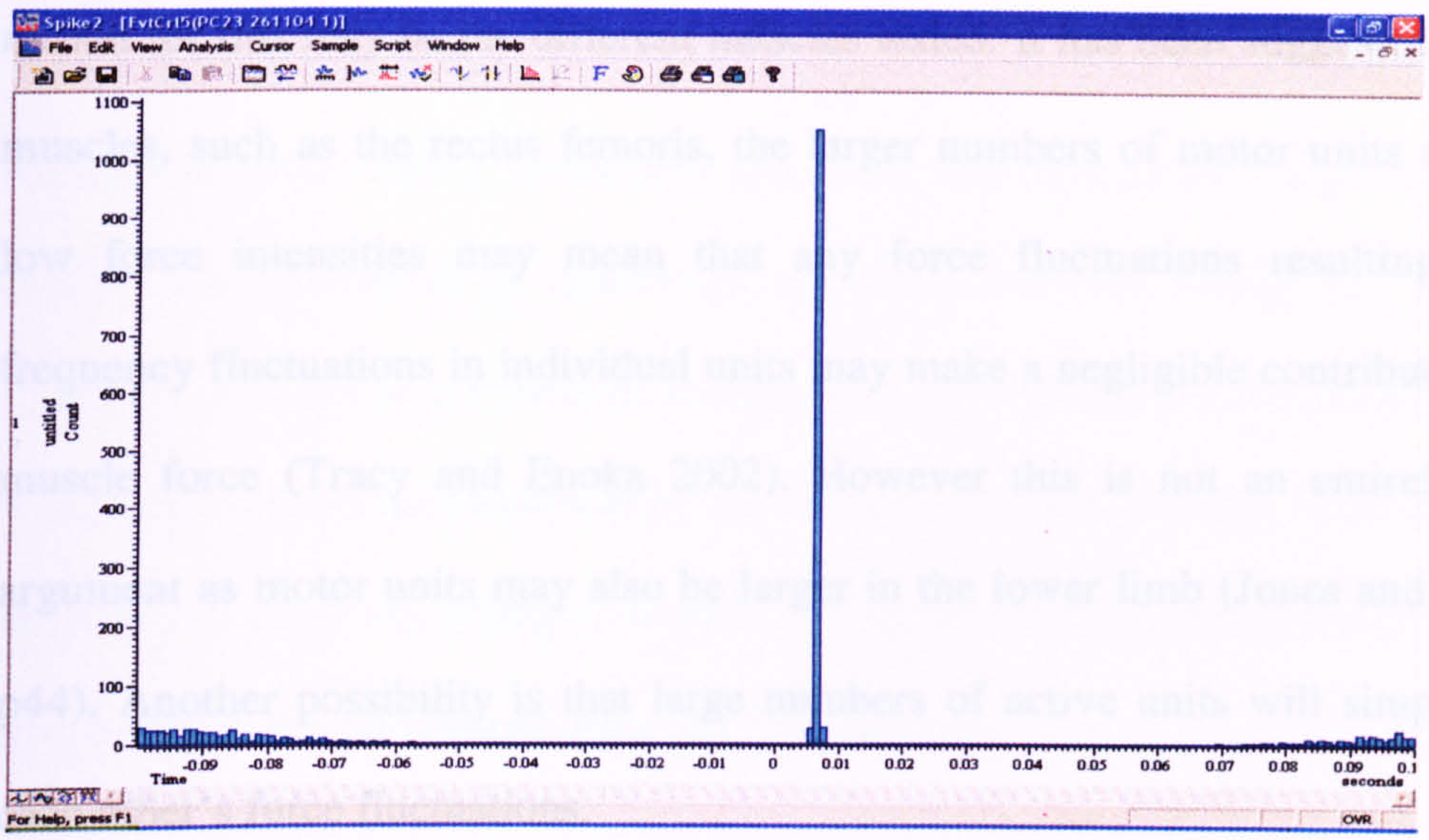


Fig. 7.8 Cross-correlelogram of motor unit discharges in an older subject. This very high peak was assumed to result from the discharges arising from the same unit.

Such extreme values were excluded from the analysis on the basis that they were unlikely to have arisen from synchrony, which normally gives peaks with CIS values much less than one sixth of these in the FDI of older subjects (Semmler et al. 2000a). They were more likely to have arisen because the two electrodes had detected signals from the same motor unit. That this was detected in 3 of the older subjects, but none of the younger subjects, suggests an age effect was responsible. It is likely that the finding was due to motor unit reorganisation, with a smaller number of larger motor units in the older subjects increasing the probability of picking up signals from the same unit.

7.4.3 Coefficient of variation of action potential firing interval

The lack of any difference between groups in the $CoV_{(ISI)}$ suggests that age differences in $CoV_{(ISI)}$ do not contribute to age-related differences in steadiness. The lack of any association was underlined by the lack of any significant correlations between all measures of steadiness and $CoV_{(ISI)}$ in either age group. This is the first study to investigate this issue in the lower limbs, but the results conflict with the findings of Laidlaw et al. (2000) in the small hand muscles, where the older subjects were found to have greater unsteadiness and a greater CoV of action potential firing frequency. One reason for this may be the different muscles tested. It has been suggested that in larger muscles, such as the rectus femoris, the larger numbers of motor units active even at low force intensities may mean that any force fluctuations resulting from firing frequency fluctuations in individual units may make a negligible contribution to overall muscle force (Tracy and Enoka 2002). However this is not an entirely convincing argument as motor units may also be larger in the lower limb (Jones and Round 1990, p44). Another possibility is that large numbers of active units will simply cancel out each other's force fluctuations.

In agreement with this study, Galganski et al. (1993) and Semmler et al. (2000a) did not detect higher discharge variability in the elderly. This may possibly be explained by the inclusion of higher force levels in their pooled analyses than Laidlaw et al. (2000). Discharge variability is most apparent at lower force levels (Erim et al. 2000, Valliancout et al. 2003, Moritz et al. 2005) and so the inclusion of higher force levels in these studies may have obscured age differences in variability. In this study, the mean frequencies in both groups of around 9 Hz are characteristic of low threshold units at low levels of force (Moritz et al. 2005) and so it is unlikely that use of high forces could explain results.

The $CoV_{(ISI)}$ did not vary with the firing frequency. Previous evidence in the rectus femoris suggests that lower firing frequencies may lead to greater variability of action potential firing when frequencies are below a threshold of 10-13 Hz (Person and Kudina 1972). This relationship probably contributes to the relationship between force levels and discharge variability mentioned above. The reason for no such relationship in this study may be because the range of frequencies was limited by the low intensity contraction; Person and Kudina (1972), in contrast, used a wider range of contraction strengths and thus elicited a greater variety of frequencies.

7.5 Conclusions

Antagonist co-activation, motor unit synchrony and motor unit firing variability do not explain the differences in quadriceps steadiness between young and old. However, synchrony is a strong correlate of steadiness in both young and old subjects. Motor unit reorganisation was demonstrated by the chance sampling of fibres from the same motor units in older but not younger subjects during needle electromyography, and this may play a role in the differences observed for steadiness.

Motor unit coherence was not analysed in this study, but previous work in the FDI (Semmler at al. 2003) suggests it may be an important factor in quadriceps steadiness changes. Further work is required in that area.

8 Relationships between different steadiness measures and function

8.1 Introduction

Several steadiness studies have concurrently measured steadiness during isometric and anisometric contractions (Burnett et al. 2000, Graves et al. 2000, Laidlaw et al. 2000, Hortobagyi et al 2001, Schiffman and Luchies 2001,Tracy and Enoka 2002) but none have evaluated the degree to which the forms of measurement correlate. Furthermore, the relationship of these measures with functional steadiness is also unknown. Knowledge of how the different measures relate to each other is potentially important, as this may shed light on whether the tests are measuring the same parameters.

Though there is an assumption in the literature that decreased steadiness may be related to poorer function in the elderly (Galganski et al. 1993, Christou and Carlton 2001) only two studies have directly investigated this. Manini et al. (2005) showed no relationship between gait speed, chair rise time and stair ascent/descent time and quadriceps isometric steadiness at 50% MVC. In contrast Tracy et al. (2004) found that improvements in anisometric steadiness after a strength training programme were accompanied by improvements in gait speed and chair rise time, although stair ascent and descent times did not change. Although it was shown that the functional improvements did not correlate with strength improvements, no correlation analysis between functional and steadiness improvements was carried out, and so the implications of this study are ambiguous.

Although unsteady movements may affect the performance of functional tasks directly, an indirect effect of steadiness on function via an effect on movement speed may be important. It has been noted that steadiness in older subjects worsens with increased speed (Christou et al. 2003a, Darling et al. 1989) and it has been suggested that older people may deliberately slow down movements to improve their poorer movement accuracy (Darling et al. 1989, Welford et al. 1984). Velocity of movement is central to performance in most aspects of function, and so such slowing would represent poorer function resulting from poorer steadiness.

Further work is necessary to establish if steadiness in the lower limbs has a significant impact on activities of daily living. In addition, if such an analysis reveals that one form of steadiness measure is more highly related to function than the others, this will endorse the use of that particular measure for investigators interested in the practical implications of steadiness. It is possible that functional steadiness may be the more appropriate measure of steadiness in this context.

There are many facets of motor function, but ambulatory function is most relevant to this study. Ambulatory function has been measured in many ways: the Timed Up and Go (TUG) test (Steffen et al. 2002), the six-minute walk (Steffen et al. 2002), sit-stand time (Suzuki et al. 2001, Skelton et al. 1994), stair climb time (Suzuki et al. 2001), stair height test (Skelton et al. 1994), 6m gait speed (Visser et al. 2000), 30m gait speed (Ringsberg et al. 1999), obstacle negotiation (Lamoureaux et al. 2003) the Berg Balance Scale (Steffen et al. 2002) and the functional reach test (FRT) (Bennie et al. 2003) are examples from the literature.

Of all tests, the TUG test appears to be the most comprehensive and relevant, encompassing the abilities of walking speed, chair rise time, turning and sitting down (Steffen et al. 2002). It has been shown to be reliable and have high construct validity (Steffen et al. 2002). It is also in widespread use in research and clinical practice (Steffen et al. 2002). It was therefore chosen as a test of elderly ambulatory function in this study. TUG test performance has been shown to be worse in elderly fallers (O'Brien et al. 1998, Shumway Cook et al. 2000, Roma et al. 2001, Rose et al. 2002, Chiu et al. 2003) and to worsen with age (Hurley et al. 1998, Steffen et al. 2002), but Daubney and Culham (1999) did not note differences between fallers and non-fallers.

Although the 6 minute walk test was designed as a measure of aerobic capacity, McNevin et al. (2002) found that it was more highly correlated to three functional tests - time to floor, time up from floor and time down stairs - than tests of strength. The six minute walk was thus also used in this study as a measure of function. Performance in the 6 minute walk and the closely related 400m walk test have been shown to be worse in fallers (de Rekeneire et al. 2003) and with age (Steffen et al. 2002).

An important determinant of ambulatory function is the ability to maintain postural equilibrium in the presence of perturbations, such as those that occur naturally during movement, i.e. balance. In addition to motor function, balance ability also depends on the patency of sensory and central components, but since all subjects studied here had been screened to exclude pathologies involving these components it can be assumed that a balance test will act as a good measure of motor function contributing to postural stability. The Berg Balance Scale is the most commonly used test, as it is highly reliable and valid, but it is also very time consuming (Bennie et al. 2003). The FRT has been shown to have similar reliability and validity (Bennie et al. 2003), but is more practical, and was therefore used in this study. The FRT has been shown to be worse in fallers (O'Brien et al. 1998, Davis et al. 1999) and with age (Duncan et al. 1990), though Daubney and Culham (1999) did not note any differences between fallers and non-fallers.

8.1.1 Implications for this study

No data exist on the relationships between different steadiness measures, and there is limited data on the relationship between steadiness and function. No previous studies have attempted to measure steadiness during functional tasks. This study aimed to evaluate any correlation of steadiness between different measures, including steadiness during functional tasks, and also to evaluate the link between steadiness and function. The hypotheses are that:

1. Steadiness measures will all correlate, but there will be a stronger relationship between the anisometric and functional measures than between the isometric measures and the others.

2. Steadiness will relate to function, but functional steadiness will have the strongest association with functional ability.

8.2 Methods

8.2.1 Subjects

Data from the whole sample of young, older fallers and older non-fallers were used (Chapter 2, page 21). All analyses in this chapter involved all subjects.

8.2.2 Tests

Data from the 3 steadiness tests documented in Chapter 5 (normalised isometric, anisometric and functional steadiness tests) were used to calculate the strength of correlation between different measures. Steadiness values of each of the normalised isometric, anisometric and functional steadiness variables were averaged between legs, as it was not wished to compare variables between different legs. The averaged values of different tests were then correlated with each other.

Three functional tests were used: the TUG, the 6 minute walk and the FRT.

Six minute walk test

Subjects were asked to trace an oval course by walking around two cones placed 8 metres apart in shoes or bare feet for 6 minutes at the fastest pace possible. The total distance covered in metres was measured. Subjects were allowed to navigate the cones in either direction.

TUG test

The subjects were seated in an armless upright chair of height 40cm. A mark was placed on the floor 3 metres away. The time taken in seconds to walk to the mark and then return to the sitting position at normal walking pace was measured.

FRT

The subjects stood straight with both arms outstretched, with the arms parallel to a wall 10cm to their right. The position of the outstretched fingers was measured on the adjacent wall. With arms still outstretched, the subject was then asked to lean forward at the spine, hip and ankle as far as possible without falling forwards, whilst maintaining the feet in the same position. The position of the outstretched fingers was again measured on the adjacent wall, and the perpendicular distance in cm between the two marks was determined.

Each of the 3 functional measurements were correlated with each of the steadiness measures on both (steady/non steady) legs, using a partial correlation technique (SPSS) correcting for bilaterally averaged peak power ($Pow_{(av)}$). This correction was performed as the functional measures correlated with $Pow_{(av)}$. ($Pow_{(av)}$ with TUG test, $R = -0.49$, $P<0.001$; $Pow_{(av)}$ with 6 minute walk test, $R= 0.72$, $P<0.001$; $Pow_{(av)}$ with FRT, $R= 0.62$, $P<0.001$) implying that in a large group with varying power, any uncorrected correlations between function and steadiness might be confounded by the effects of varying power. Power has also been shown to be a strong correlate of function in healthy older people in other studies (Suzuki et al. 2001, Bean et al. 2002).

The other main determinant of function in healthy older people appears to be strength (Brown et al. 1995, Ringsberg et al. 1999, Suzuki et al. 2001, Sobolewski et al. 2001). For the main analysis, $Pow_{(av)}$ was used as a covariate in preference to bilaterally averaged peak strength ($St_{(av)}$) as strength correlated with the functional measures less strongly ($St_{(av)}$ with TUG test, $R = -0.451$, $P<0.001$; $St_{(av)}$ with 6 minute walk test, $R= 0.641$, $P<0.001$; $St_{(av)}$ with FRT, $R= 0.477$, $P<0.001$). However, a secondary test involving correction for strength was carried out.

8.3 Results

8.3.1 Group characteristics

The group characteristics were as described in tables 3.1 (page 45) and 4.1 (page 85)

8.3.2 Correlations between different steadiness measures

Within each type of steadiness measure (i.e. normalised isometric) all correlations were highly significant (P<0.001), with R values varying between 0.31 and 0.74. All significant correlations were positive. The three types of steadiness test appeared to have similar intra-test associations (Table 8.1, tan portions).

Between diferent types of steadiness measure (i.e. between isometric and functional steadiness variables) there were less significant and less consistent correlations between variables. For significant associations, R values were modest, ranging from 0.21 to 0.42, and all were positive. The isometric and functional steadiness measures correlated with each other better than with other measures, with 3/12 significant combinations at P<0.05, and a further 2 at P<0.01 (Table 8.1, green portion). The anisometric and isometric measures showed only 2/12 significant combinations at P<0.05, and a further 1 at P<0.01 (Table 8.1, yellow portion) whilst the functional and anisometric measures showed only 1/16 significant correlations at P<0.05 (Fig. 8.1, blue portion).

	25% MVC	10% MVC	STAND	SIT	STEP UP	STEP DOWN	CON1	ECC1	CON5	ECC5
50% MVC	0.46^	0.31^	-	-	0.30*	0.33*	0.33+	-	-	-
25% MVC		0.72^	-	-	0.25+	-	-	0.42*	-	-
10% MVC			0.21+	-	0.23+	-	-	0.33+	-	-
STAND				0.58^	0.46^	0.47^	-	-	-	-
SIT					0.32^	0.50^	-	-	-	-
STEP UP						0.74^	-	-	-	-
STEP DOWN							0.28+	-	-	-
CON1								0.67^	0.47^	0.70^
ECC1									0.50^	0.70^
CON5										0.50^

Table 8.1. Correlations (R values) between different steadiness measures for the whole sample. +P<0.05, *P<0.01, ^P<0.001. Only significant correlations are shown. (n = 56-110). Intra-measure correlations are shown in the tan areas. Inter-measure correlations are shown in yellow (isometric v anisometric), green (isometric v functional) and blue (anisometric v functional) CON= concentric, ECC= eccentric, 1= 1kg, 5= 5kg.

8.3.3 Group differences in function

The young performed better than the older non-fallers, and the non-fallers performed better than the fallers, in all three tests (Table 8.2). The six minute walk interacted with sex (with male subjects having higher values), and the analysis was corrected for this.

	Young		Fallers		Non-fallers	
	n	mean (SE)	n	mean (SE)	n	mean (SE)
TUG (sec)	40	7.2 (0.2)*	31	11.7 (0.6)*	42	9.2 (0.4)
FRT (cm)	41	39 (1.2)*	31	26 (1.4)*	42	31 (1.1)
6 min walk (metres)	40	596 (16) *	30	361 (18)+	40	445 (17)

Table 8.2 Function in young, fallers and non-fallers. * significantly different to non-fallers ($P<0.01$), + significantly different to non-fallers ($P<0.05$)

8.3.4 Correlation of steadiness with function

When correcting for bilaterally averaged power, there were significant positive associations between the time taken to complete the TUG test and the standard deviation of acceleration fluctuations during lower load concentric and eccentric contractions. However, The TUG test did not correlate with other steadiness variables, and the FRT and 6 minute walk test did not correlate with any functional variables (Table 8.3).

In the secondary analysis, where correlations were corrected for bilaterally averaged strength, time taken to complete the TUG test positively correlated with steadiness during the isometric contractions at 50 and 25% MVC on the steadier legs, the 1kg concentric and eccentric contractions, and the step down on the steadiest leg. The distance walked during the 6 minute walk test correlated negatively with steadiness during the isometric contractions at 50 and 25% MVC on the steadier leg, and 10% MVC on the less steady leg, and the concentric and eccentric 1kg contractions. The negative correlation indicates that larger distances covered were associated with lower force fluctuations. The FRT did not correlate with any steadiness measures (Table 8.4).

Steadiness variable		TUG (R)	6 min walk test (R)	FRT (R)
50% MVC	steadier	-	-	-
	less steady	-	-	-
25% MVC	steadier	-	-	-
	less steady	-	-	-
10% MVC	steadier	-	-	-
	less steady		-	-
Step up	steadier	-	-	-
	less steady	-	-	-
Step down	steadier	-	-	-
	less steady	-	-	-
Stand	steadier	-	-	-
	less steady	-	-	-
Sit	steadier	-	-	-
	less steady	-	-	-
1kg	concentric	0.32+	-	-
	eccentric	0.33+	-	-
5kg	concentric	-	-	-
	eccentric	-	-	-

Table 8.3. Correlations between steadiness measures and TUG test, 6 minute walk test and FRT for the whole sample (n = 47-87). All correlations are corrected for leg power (bilateral average). +P<0.05, only significant correlations are shown.

Steadiness variable		TUG (R)	6 min walk test (R)	FRT (R)
50% MVC	steadier	0.24+	-0.31+	-
	less steady	-	-	-
25% MVC	steadier	0.27+	-0.29+	-
	less steady	-	-	-
10% MVC	steadier	-	-	-
	less steady		-0.25+	-
Step up	steadier	-	-	-
	less steady	-	-	-
Step down	steadier	0.23+	-	-
	less steady	-	-	-
Stand	steadier	-	-	-
	less steady	-	-	-
Sit	steadier	-	-	-
	less steady	-	-	-
1kg	concentric	0.36+	-0.35+	-
	eccentric	0.38+	-0.34+	-
5kg	concentric	-	-	-
	eccentric	-	-	-

Table 8.4. Correlations between steadiness measures and TUG test, 6 minute walk test and FRT for the whole sample (n = 46-87). All correlations are corrected for quadriceps strength at 80° (bilateral average). +P<0.05, only significant correlations are shown.

8.4 Discussion

8.4.1 Correlation between measures of steadiness

The highly significant correlations between variables within each category of test (intra-test correlations) suggest each test was measuring a similar parameter to other tests within the same category. In other words, this showed that steadiness at high forces was related to low force steadiness, or that concentric and eccentric steadiness tended to be related when measured with the same type of test. This indicates that unsteadiness at high and low force levels, and in shortening/lengthening contractions may have similar mechanisms. In contrast, the number and strength of correlations between tests from different category variables (inter-test correlations) were smaller. These will be referred to as inter-test correlations.

The significant inter-test correlations were generally weak, with largest R^2 values of only 0.18. However, steadiness during the three isometric steadiness measures correlated consistently, albeit weakly, with steadiness during the step up. This suggests that the isometric and step up tests may have been measuring similar parameters, despite the step up test also involving ankle and hip muscles. The less consistent associations between isometric steadiness tests and the standing and sitting tasks may have been because the standing and sitting movements involved a greater contribution from the hip muscles than the step up movement; hence a correlation with the isometric contractions utilising only the quadriceps was less likely. However, this would not explain the less consistent correlation between the step down and isometric measures, although the single correlation between steadiness during the the step down and the 50% MVC isometric contraction had the highest R value among all correlations.

The lower intensity isometric measures also correlated consistently with steadiness during the lower intensity eccentric contraction. The association between isometric and eccentric but not concentric steadiness measures is difficult to explain, and conflicts with the concentric step task having the better association with isometric measures. The lack of correlation between isometric and higher load anisometric measures of steadiness is also difficult to explain. Although the greater inertia of the heavier concentric load might lead to greater gross volitional accelerations which could influence non-volitional acceleration fluctuations, the acceleration traces showed no signs of low frequency volitional accelerations imposed on the higher frequency non-volitional accelerations for either load (see Figs. 5.9-10, pages 139-140).

It had been hypothesised that the anisometric and functional steadiness tests would have the closest association due to their dynamism and their common measurement of standard deviation of acceleration. Hence the relatively poor association between them was surprising. The different loads involved in the two tests may be part of the explanation, with the functional tests involving a large percentage of body weight and the anisometric tests involving a very low percentage. Another explanation may have been that the anisometric tests involved only the knee extensors but the functional tests involved additional muscle activity from hip and ankle flexors and extensors. However both these explanations conflict with the associations seen between steadiness during higher intensity isometric quadriceps contractions and during stepping up and down.

8.4.2 Function across groups

As expected, the TUG, FRT and 6 minute walk were better in young than older subjects, and also better in non-fallers than fallers. This concurs with most previous work (Duncan et al. 1990, Hurley et al. 1998, O'Brien et al. 1998, Davis et al. 1999, Steffen et al. 2002, de Rekeneire et al. 2003, Shumway Cook et al. 2000, Roma et al. 2001, Rose et al. 2002, Steffen et al. 2002, Chiu et al. 2003).

8.4.3 Correlation between steadiness and function, when correcting for power

Allowing for the intervening effects of power, unsteadiness during lower load eccentric and concentric contractions increased with the slowness of the TUG test. Hence poorer anisometric steadiness does appear to have an association with lower functional level when power is controlled for. However, isometric and functional streadiness measures did not appear to relate to TUG performance. In addition, the other functional tests did not relate to any steadiness variable.

In the only other study to analyse associations between function and steadiness, Manini et al. (2005) used similar functional indices to this study, but only used isometric steadiness (normalised force fluctuations at 50% MVC) as a steadiness variable. As in this study, they did not note any associations between isometric steadiness and function.

A causal relationship between anisometric steadiness and function cannot be assumed. It is possible that anisometric steadiness could correlate with function by virtue of also correlating with another factor that had a causal relationship with function. However, given the fact that power may be the major determinant of function in the healthy elderly (Suzuki et al. 2001, Bean et al. 2002), and that power was corrected for, a causal relationship looks plausible. A mechanism for decreased steadiness causing poorer functional performance is hinted at by Darling and colleagues' (1989) suggestion that older people may slow down their movements to optimise their worse steadiness.

The strength of the association between anisometric steadiness and function was modest, with R^2 values for the significant associations of 0.10-0.11. Hence variations in steadiness will only contribute slightly to changes in function. However, steadiness during the low load eccentric contraction also related to falling (chapter 6), suggesting that this measurement may be of clinical value.

The association between function and the anisometric but not isometric steadiness variables is unsurprising, as a dynamic steadiness measure might be expected to relate better to dynamic function. However, for the same reasons, the lack of association between function and the functional steadiness measures is surprising. The standing and sitting functional steadiness tasks involved similar movements to some of those involved in the TUG test. Moreover, the functional steadiness tests measured the aggregate of steadiness at the ankle, knee and hip, which are invariably in a weight-bearing position during lower limb functional activities. One possibility is that the greater complexity of the functional steadiness tests led to measurements which did not reflect steadiness as validly as for the other anisometric test. This might then explain the functional steadiness tests' relative lack of correlation with function. However, the fact that the functional steadiness tests correlated relatively well with the other measures makes this less likely. Nevertheless, consideration of the validity of the three different types of steadiness test is warranted, and this will be discussed in section 8.4.5

8.4.4 Correlation between steadiness and function, when correcting for strength

When correction was performed for bilaterally averaged strength instead, many more significant associations were observed. Unsteadiness during isometric, eccentric and concentric contractions, and during stepping down, increased with the slowness of the TUG test. Similarly, isometric, concentric and eccentric unsteadiness increased with the slowness of the 6 minute walk test. Moreover, the strengths of association were slightly greater, with R^2 values of 0.14 for the low load eccentric steadiness measure. Three isometric and eccentric variables showing an association with function - isometric steadiness at 25% in the steadier leg and 10% in the less steady leg, and eccentric steadiness during the 1kg contraction – were also associated with falling (Chapter 6) suggesting that these are important clinical variables.

Whilst these results are of interest, they are probably less valid than those obtained when power was corrected for, as power was shown to correlate more strongly with the functional indices, particularly the FRT and the 6 minute walk, and this has also been shown by other workers (Bassey et al. 1992, Skelton et al. 1994, Foldvari et al. 2000, Suzuki et al. 2001, Bean et al. 2002).

It is revealing that using maximal power as a covariate should eliminate the significant associations between function and steadiness during isometric contractions and stepping down that were evident when using maximal strength as a covariate. The difference between maximal strength and maximal power is that the latter encompasses the speed of movement in addition to strength, and it may therefore be that speed rather than isometric or functional steadiness is the crucial factor governing function; if correcting for a covariate (ie speed) eliminates an effect, the covariate may well be the cause of the effect itself. Importantly, this speed, which relates to maximal speed, may be unrelated to the level of functional speed dictated (if at all) by steadiness. Hence the impression that function and isometric or functional steadiness are related should be interpreted with some caution. However, the results obtained using power as a covariate suggest function relates to low load concentric and eccentric steadiness in addition to maximal speed, and thus these steadiness measures can be validly regarded as likely independent contributors to function.

8.4.5 Validity of steadiness measures

As there is no gold standard measure of steadiness that could be applied to all types of test, it is not possible to establish criterion validity for any of the tests. Construct validity, however, can be assessed. Construct validity is determined by the extent to which the theoretical constructs upon which the test is based are valid (Everitt 2003 p53) and the following discussion will attempt to evaluate this.

The isometric test is a very simple way of measuring steadiness. If steadiness is viewed as the degree of fluctuations of force, then this method, which measures force via a strain gauge, is direct. The use of visual feedback has already been discussed as a possible threat to validity (Chapter 5), if we define steadiness as being a form of force control that does not normally depend on visual feedback of the force levels. However, as discussed in chapter 5, visual feedback is a practical necessity when measuring steadiness at discrete contraction intensities. Another threat to the construct validity is damping effects from the measurement system. Damping effects in such a mechanical system do not seem to have been evaluated, but it is to be expected that amplitude attenuation of specific frequencies of acceleration fluctuations might occur.

The anisometric test measures the fluctuations in linear acceleration in the moving cord, which is attached to each subject's shin and is used to lift the load. Construct validity is established if it can be shown that these fluctuations in acceleration are likely to be proportional to the fluctuations in knee extensor force output. One threat to such proportionality comes from accelerations in the cord that are due to the gross volitional acceleration and decceleration of the load. However, as previously explained, volitional accelerations were negligible compared to the non-volitional accelerations. Varying frictional resistance between the cord and pulleys had the potential to create spurious cord accelerations, but low friction pulleys and cord were used to minimise this effect. It is possible that damping may have been a factor in incomplete transcription of muscle force fluctuations to linear cord accelerations.

The functional steadiness tests measure the fluctuations in angular knee acceleration during the functional movements. Again, acceptable construct validity requires that these fluctuations in angular acceleration are proportional to the fluctuations in knee extensor (and gluteal) force output. One threat to such proportionality comes from angular accelerations that are due to the gross volitional acceleration and decceleration of the movement. This could have been avoided by only measuring fluctuations during the middle part of the movement. However, in many cases, this would have meant sampling fluctuations over a duration of < 0.5 seconds, which would have permitted sampling of very few acceleration fluctuations. Observation of the traces showed that these volitional gross accelerations were negligible compared to the non-volitional accelerations. In most cases, no spikes of acceleration were seen at the points where angular velocity changed from positive to negative or vice versa, which were the points at which volitional acceleration and deceleration should have been at their most intense. Importantly, the acceleration power at low frequencies of 1-4Hz was relatively low for all groups, indicating that the contribution of volitional accelerations was not important.

Another threat to validity was the positioning of the markers during motion analysis. Although the hip and ankle positions would not be critical, the knee marker was required to be at the centre of rotation to avoid distortion of the angular measurements. Great care was taken to ensure this, but perfect placement was always unlikely. The stability of markers was also crucial, as any vibrations or skin movements would affect angular measurements. Firm attachment of the markers to the skin, and ensuring that marker leads were unimpeded probably minimised this source of error.

An important threat to proportionality could be the damping effects from the heavy body segments. These may possibly have been a more powerful confounding factor in the functional than anisometric tests due to the greater resistance provided by the whole body weight. However, given the similar body weights between groups, this may not have confounded results greatly.

It is possible that some of the acceleration fluctuations could be due to postural corrections. This could confound results, as older subjects (Okuzumi et al. 1996, Skalska et al. 2004), especially fallers (Lord et al. 1999, Melżer et al. 2004), may have greater postural sway. However, such sway involves the whole body, and is more pronounced in the medio-lateral direction in older people (Okuzumi et al. 1996) and fallers (Lord et al. 1999, Melzer et al. 2004). Hence significant effects on saggital movements at the knee are unlikely.

The hand-support pole provided for the stepping tasks was solely to assist balance, so validity of the stepping tests could be reduced by subjects using it as a propulsive aid when stepping up or a controlling aid when stepping down. All subjects were informed of the importance of a light grip on the pole, and any trials where the subject had clearly used the pole for advantage were excluded. Manifestations of excessive use of the pole were obvious contraction of upper limb and hand muscles.

This discussion suggests no clear difference in construct validity between the three types of test, although the lack of data on damping effects does not permit a conclusive conclusion. Researchers are advised to use the functional tests as described, but with some modifications. To eliminate the threat of upper limb involvement in the stepping tasks, future work could measure EMG at the shoulders to detect excessive use of the arms, or replace the support pole with a safety harness designed to provide support only if a fall occurs. Stepping down should occur forwards to maximise relevance to daily activities.

8.5 Conclusions

Different measures of steadiness do not correlate with each other strongly or consistently, indicating that they may not be measuring the same parameters. Better quadriceps steadiness during anisometric contractions correlates with better ambulatory function. Construct validity of the three tests is probably adequate.

9 Muscle strength training in the elderly: effects on steadiness, strength, power and muscle CSA.

9.1 Introduction

9.1.1 Training and steadiness

9.1.1.i Strength training effects on isometric steadiness

Although strength training has led to isometric steadiness improvements in the FDI (Laidlaw et al. 1999, Keen et al. 1994) it has failed to elicit isometric steadiness improvements in the quadriceps (Tracy et al. 2001, Hortobagyi et al. 2001, Bellew et al. 2002, Tracy et al. 2004). Tracy et al. (2004) did detect an improvement in isometric steadiness at 50% MVC in response to training, but this was not different to that in a control group. High loads at around 80% of a one repetition maximum (1RM) (Laidlaw et al. 1999, Hortobagyi et al. 2001, Bellew et al. 2002, Tracy et al. 2004, Tracy et al. 2001) and those as low as 30% 1RM (Hortobagyi et al. 2001) and 10% 1RM (Laidlaw et al. 1999) produced similar effects. Although these results suggest that isometric quadriceps steadiness does not respond to strength training, none of these interventions exceeded 16 weeks and it is possible that longer term interventions might produce different results.

9.1.1.ii Strength training effects on anisometric steadiness

Hortobagyi et al. (2001) and Tracy et al. (2004) demonstrated training induced improvements in both concentric and eccentric steadiness in the quadriceps, whilst Laidlaw et al. (1999) demonstrated similar effects in hand muscles. Hortobagyi et al. (2001) and Laidlaw et al. (1999) used high and low strength training loads, with similar effects. Hortobagyi et al. (2001) did not take account of the increased strength of the exercise groups after training when measuring steadiness. An absolute force target level of 25N was used for all subjects, so the stronger trained subjects were working at a lower percentage of MVC than they had initially, which may have confounded results. In addition, Tracy et al. (2004) did not observe an effect of training on concentric and eccentric steadiness at target levels < 50% 1RM. Hence the evidence for strength effects on anisometric quadriceps steadiness is not strong and further work is required.

9.1.1.iii Strength training effects on functional steadiness

No previous work has addressed the effects of any form of training on steadiness measured during functional tasks.

9.1.1.iv Other training techniques

Some studies tested the effect of matching training movements with a visual joint angle template, referring to this as steadiness training (Patten and Kamen 2000, Tracy et al. 2001, Tracy et al. 2004). This was found to improve quadriceps anisometric, but not isometric, steadiness (Tracy et al. 2004), and dorsiflexor isometric steadiness (Patten and Kamen 2000) although the latter did not use a control group. Tracy et al. (2004) and Patten and Kamen (2000) used loads of 60-80% 1RM so it is not possible to extricate steadiness and strength training effects. Tracy et al. (2001) used both low and high load steadiness training in the quadriceps, but neither were effective in reducing steadiness.

Ranganathan et al. (2001b) reported that skill training, involving rolling two small metal balls around the dominant hand, improved isometric steadiness of a pinch grip. Tai Chi has also been shown to improve isometric quadriceps steadiness in the quadriceps (Christou et al. 2003b) and vertical pressure force fluctuations in the arm (Yan et al. 1999).

9.1.1.v Effect of faller status on steadiness changes

Since fallers may have worse steadiness it is important to establish if they can improve steadiness in response to strength training. This has not previously been studied.

9.1.1.vi Asymmetry of steadiness

No reports were found on the effects of training on steadiness asymmetry.

9.1.1.vii Mechanisms for training effects on steadiness

In many studies, a direct strength training effect does not appear to have been responsible for any steadiness improvements. For example, Laidlaw et al. (1999) and Hortobagyi et al. (2001) obtained similar results in both low and high intensity training groups. Moreover, strength gains were sometimes not observed despite improvements in steadiness (Laidlaw et al. 1999, Keen et al. 2004). These results are unsurprising given the lack of a consistent relationship between strength and steadiness in older subjects (Chapter 6). One explanation given by Laidlaw et al. (1999) for the improvements in steadiness was a neural 'transfer effect' from the movements carried out in the training programmes. This mechanism has also been suggested for the effects of Tai Chi (Christou et al. 2003b, Yan et al. 1999).

Laidlaw et al. (1999) noted that improvements in FDI steadiness were accompanied by reductions in antagonist co-activation and Patten and Kamen (2000) demonstrated a similar effect in the ankle. However no correlation analysis was performed in either study. Moreover, the lack of a control group in the latter study means that practice effects cannot be excluded as reasons for the improved steadiness.

Semmler and Nordstrom (1998) found that weightlifters had greater FDI motor unit synchrony and lower isometric FDI steadiness than musicians, which initially suggests that strength training may increase force variability due to greater motor unit synchrony (Barry and Carson 2004). Similarly, weightlifters were also shown to have greater motor unit coherence (Semmler at al. 2004) which may also lead to lower steadiness (Taylor et al. 2003). However, weightlifting does not necessarily lead to training effects in the FDI, so this does not imply that direct training of a muscle affects motor unit synchrony or coherence (and thus steadiness) in this way.

9.1.2 Effects of training on lower limb strength and muscle area in older people

There have been many controlled studies showing that strength training improves strength in the quadriceps (Fiatarone et al. 1994, Pyka et al. 1994, Lord et al. 1995, Lexell et al. 1995, Morganti et al. 1995, Skelton et al. 1995, Sipila et al. 1996, Wolfson et al. 1996, Meuleman et al. 2000, O'Neill et al. 2000, Adams et al. 2001, Hakkinen et al. 2001, Hortobagyi et al. 2001, Miszko et al. 2003, Lamoureux et al. 2003, Seynnes et al. 2004, Capodiaglo et al. 2005, De Vreede et al. 2005, Reeves et al. 2005), the hamstrings (Lord et al. 1995, Cress et al. 1999, Lamoureux et al. 2003) and the quadriceps and hamstrings combined (Robinson et al. 2004) in older people. In contrast, MacRae et al. (1994) did not observe any quadriceps strength improvements isometrically, but the intervention utilised only exercises against body weight (i.e. step ups) rather than traditional weight training. Meuleman et al. (2000) also did not show any concentric improvements in nursing home residents.

Although some of these studies measured strength using the 1-3RM of the training exercise (Pyka et al. 1994, Fiatarone et al. 1994, Morganti et al. 1994, Adams et al. 2001, Lamoureux et al. 2003, Miszko et al. 2003), thus possibly measuring improvements in skill rather than strength, the majority used isometric (Sipila et al. 1996, Skelton et al. 1995, Hakkinen et al. 2001, Lord et al. 1995, Meuleman et al. 2000, Hortobagyi et al. 2001, Robinson et al. 2004, Seynnes et al. 2004, De Vreede et al. 2005) or concentric (Lexell et al. 1995, Wolfson et al. 1996, O'Neill et al. 2000, Hortobagyi et al. 2001, Hakkinen et al. 2001, Cress et al. 1999, Capodiaglo et al. 2005, Reeves et al. 2005) measurements. Only Hortobagyi et al. (2001) and Reeves et al. (2005) have examined eccentric strength, and only the former noted improvements.

These results are supported by several studies showing the positive effects of strength training on muscle size in older people (Frontera et al. 1988, Fiatarone 1990, Roman et al. 1993, Pyka et al. 1994, Fiatarone 1994, Lexell et al. 1995, Welle et al. 1996, Harridge et al. 1999, Tracy et al. 1999, Ferri et al. 2003, Reeves et al. 2004b). These findings emphasise that strength gains are morphological as well as neural. Although only 5 have used controls (Fiatarone et al. 1994, Pyka et al. 1994, Lexell et al. 1995, Tracy et al. 1999, Reeves et al. 2004b) the nature of this measurement means that values after training are unlikely to be artificially elevated by subject practice effects. In contrast, Sipila and Suominen (1996) did not note any changes in CSA, and they are the only study to use ultrasonography. In conclusion, strength training appears to elicit hypertrophy and strength gains in the knee flexors and extensors in older people, although the effects on eccentric strength are unconvincing.

Although hypertrophy appears to accompany strength increases, there is also evidence that strength training can improve F/CSA or Force/volume (Welle et al. 1996, Tracy et al. 1999, Reeves et al. 2004b) through improved activation (Scaglioni and Ferri 2002, Reeves et al. 2004b, 2004c, 2005) and reduced antagonist co-contraction (Hakkinen et al. 1998b, Hakkinen et al. 2001). However, some studies have not noted changes in F/CSA (Harridge et al. 1999), voluntary activation (Skelton et al. 1995, Harridge et al. 1999) or co-contraction (Reeves et al. 2003b, 2004b, 2004c, 2005).

In controlled studies, Tai Chi training has also been reported to lead to increases in quadriceps (Lan et al. 1998, Christou et al. 2003b) and hamstring (Lan et al. 1998) strength. Wolfson et al. (1996) also showed that long term Tai Chi training led to maintenance of strength gains elicited by strength training. Tai Chi combined with strength training has also been claimed to be effective but no comparison was made with a control or other groups (Capodiaglio et al. 2005). Christou et al. (2003b) suggested that Tai Chi may improve strength through improved co-ordination, whilst Wolfson et al. (1996) suggested that the flexed hip and knee postures in many Tai Chi movements may have a strength training effect. Tai Chi may be a useful adjunct to strength training, and studies are required where pure strength training is compared to a combined Tai Chi and strength training regimen.

9.1.3 Effects of strength training on lower limb power in older people

Specific power training utilising high velocity contactions has been shown to increase power relative to a control group (Earles et al. 2001, Hruda et al. 2003, Newton et al (2002), Hakkinen and Hakkinen 1995) and also to a traditional 8RM strength training group (Fielding et al. 2002). However there have been conflicting findings on both counts (Miszko et al. 2003) and comparisons of the efficacy of high velocity and traditional regimes have been biased by both regimes working at the same moderate intensity of 70% 1RM (Fielding et al. 2002) which may be suboptimal for strength training. In addition, three of these studies included slow velocity contractions in the regimen (Hruda et al. 2003, Newton et al. 2002, Hakkinen and Hakkinen 1995). It has been questioned whether older people can tolerate high velocity training as well as younger people (Newton et al. 2002). There is therefore a case for evaluating the effect of traditional low velocity strength training on power.

Although several studies have claimed positive effects of low velocity strength training on lower limb power in older people (Fiatarone et al. 1994, Jozsi et al. 1999, O'Neill et al. 2000, Hakkinen et al. 2001, Izquierdo et al. 2001, Seynnes et al. 2002, Fielding et al. 2002, Ferri et al. 2003, De Vreede et al. 2005, Capodiaglo et al. 2005) only Seynnes et al. (2002), Capodiaglo et al. (2005) and De Vreede et al. (2005) included a control group. O'Neill et al. (2000) used a control leg and Hakkinen et al. (2001) used an initial control period. Three conflicting controlled studies exist (Skelton et al. 1995, Adams et al. 2001, Miskzo et al. 2003) though Skelton et al. (1995) used elastic tubing training resistances instead of weight training, which may not provide adequate resistance.

9.1.4 Training to reduce strength and power asymmetry

No studies were found that investigated whether strength training can reduce asymmetry of strength, power or muscle CSA. Asymmetry is a possible risk factor for falls (Chapter 4, Skelton et al. 2002).

9.1.5 Effects of faller status on strength, power and muscle size changes

Only one study has investigated the effect of faller status on strength training effects (Robinson et al. 2004) suggesting that both fallers and non-fallers show similar increases in strength with training.

9.1.6 Implications for this study

The effects of longer-term strength training on isometric and anisometric steadiness remain to be identified. In particular, the evidence for anisometric steadiness changes with exercise is limited and no studies have investigated the effects of training on functional steadiness. Co-activation changes may be a mechanism for steadiness changes with training, but the evidence is limited. Controlled studies have shown conflicting evidence about the effects of low velocity training on lower limb power and eccentric strength. No studies have investigated the effects of training on asymmetry of strength and power, or the effect of faller status on steadiness, strength or power changes. The hypotheses of this chapter are:

1. Longer term strength training leads to improvements in isometric, anisometric and functional steadiness

2. Co-activation changes correlate with steadiness changes

3. Strength training improves power, eccentric strength and reduces asymmetry of strength and power.

4. Fallers and non-fallers do not differ in their response to training

5. A combined Tai Chi and strength training regimen may improve steadiness and strength more than a pure strength training regimen.

9.2 Methods

9.2.1 Subjects

Subjects were derived from the elderly faller and elderly non-faller groups described in Chapter 2. The young group were not included.

9.2.2 Tests

The isometric and isokinetic strength, power and rectus femoris CSA measurements (Chapters 3 and 4); isometric, anisometric and functional steadiness measurements (Chapters 5 and 6); and muscle co-activation measurements during functional activities (Chapter7) were repeated at the end of a 12 month controlled intervention study. The protocol was similar to baseline testing, with the exceptions that only quadriceps and hamstring isometric and isokinetic strength were measured, and isometric angles were limited to 90,80,60,50 and 30° knee flexion.

9.2.3 Intervention

Subjects were assigned to either an intervention group or control group by choice. Randomisation was not used in order to maximise recruitment, particularly for geographical reasons. Their decision was adhered to as far as possible. Thirty eight subjects entered the control group (23 non-fallers, 15 fallers) and 40 subjects (21 non-fallers, 19 fallers) entered the intervention group. Eleven withdrew from the control group (6 non-fallers, 5 fallers), and three withdrew from the exercise group (all fallers). All withdrawals were for reasons of ill health. All those withdrawing did not attend the final 12 month test.

9.2.3.i.i *Control group*

Subjects in the control group were not given any intervention, but were expected to continue with their normal routine. No restrictions were placed on their activities.

9.2.3.i.ii Intervention group

The exercise intervention involved 12 months of twice-weekly exercise sessions taken in a group of up to 10 subjects, supervised by an accredited exercise instructor. Five instructors supervised sessions in 7 venues in London. These sessions lasted 1 hour and consisted of:

- A 20 minute warm up including stretching and low level aerobic activities to raise the heart-rate to <70% of the maximum (which was predicted by 220 beats. minute^{-1} minus their age). This was not intended to have an aerobic training effect.

- Muscle strengthening work, involving 2-3 sets of 8-10 repetitions at 80% of 1 repetition maximum (1RM) of the quadriceps and hamstrings, using appropriate multi-gym equipment (usually leg press, quadriceps extension, hamstring curl). Both legs were trained, but single leg contractions were used, and the load used was specific to each leg. Upper limb work was included at the instructor's discretion. In the first 2 months of training the training load was gradually raised from 20% to 80% of 1RM.

- A warm down, including a brief session of Tai Chi. Tai Chi consists of 108 precise movements of the body whilst standing, utilising a variety of muscles and body positions, although only a small selection of these were used at the instructor's discretion (Wolf et al. 1996). Tai Chi was used in only 19 of the 40 exercise subjects.

Two sessions of home exercise were also prescribed. Each included a 20 minute quadriceps and hamstring strengthening session using elastic resistance (theraband). Subjects were issued with an instruction booklet and taught the exercises in the classes. Each venue was equipped with at least one leg press, at least one leg-curl, at least two exercise bicycles and sufficient space for floor-work.

9.2.4 Data analysis

For post-test data analysis, legs were defined as steady/less steady or strong/weak according to pre-test values. This was to ensure that pre to post-test comparisons were always performed on the same leg.

Repeated measures ANOVAs with the group as the between-subjects factor and time as the within-subjects factor were used for most analyses. These are paired analyses, so data from subjects not performing the post-tests was not included.

A preliminary repeated measures ANOVA was initially performed for all variables to assess the intervention group (strength and Tai Chi *vs* strength only) x time (Pre to post-test) interaction; that is whether the two intervention groups differed in changes over time from pre to post-test. The cofactors are described in the results section. For those variables not showing an intervention group x time interaction, the intervention groups were pooled to a single exercise group (pooled exercise) to optimise statistical power.

Pooled analysis of steadiness, strength, power, RF CSA and asymmetry

For all variables that did not show an intervention group x time interaction in the previous analysis, the main GLM repeated measures ANOVA was performed to detect group (pooled exercise *vs* control) x time interactions. The cofactors are described in the results section.

Non-pooled analysis of steadiness, strength, power, RF CSA and asymmetry

For the variables that showed a significant intervention group x time interaction, a GLM repeated measures ANOVA was performed to detect group (strength only *vs* strength and Tai Chi *vs* control) x time interactions. A Bonferroni post hoc test was used as appropriate. Cofactors are described in the results section.

Analysis of co-activation and steadiness association

Percentage changes from pre- to post-test for hamstring to quadriceps activation ratios (H/Q) and plantarflexor to dorsiflexor activation ratios (PF/DF) during the functional tasks were correlated with steadiness percentage changes from pre- to post-test.

9.3 Results

9.3.1 Subject characteristics.

Due to the loss of several subjects the characteristics of the exercise and control groups changed from baseline, and the new characteristics were compared across exercise and control groups with a GLM univariate ANOVA with sex as a cofactor (Table 9.1). The groups did not differ in age, weight, height, baseline activity levels, age-corrected height or age-corrected height squared. The groups differed in the ratio of men and women and fallers and non-fallers.

	Exercise		Control		P	
	N	Mean(SE)	N	Mean(SE)		sig. cofactors
Age	35	75.7 (0.63)	29	76.3 (0.92)	>0.05	-
Body mass(kg)	35	70 (1.7)	29	69 (1. 8)	>0.05	sex
Height(m)	35	1.66 (0.01)	29	1.68 (0.01)	>0.05	sex
Corr Height(m)	35	1.70 (0.01)	29	1.72 (0.01)	>0.05	sex
Corr Height^2 (m)	35	2.91 (0.04)	29	2.98 (0.04)	>0.05	sex
Baseline activity levels $	21	30.4 (3.2)	11	26.0 (4.3)	>0.05	sex
Men(n)	8		7		NA	
Women(n)	27		22		NA	
Fallers(n)	15		11		NA	
Non-fallers(n)	20		18		NA	

Table 9.1 Subject characteristics in the exercise and control groups. The groups differed in ratios of sex and faller status. $ = minutes per day at >moderate activity level (>1952 counts/min on actigraph monitor).

9.3.2 Cofactors in the analyses

Sex x time (i.e. the effects of sex on the changes in the variable from pre- to post-test) and faller status x time (i.e. the effects of faller status on the changes in the variable from pre- to post-test) were used as co-factors, given that it is unknown whether sex and faller status affect steadiness changes in response to training. The interaction of group x faller status x time was also included to show if the following different subgroups differed in their subgroup x time interaction: exercise group fallers, exercise group non-fallers, control group fallers, control group non-fallers. Other interactions were not used as this would have led to very small subgroups (n=1). Step-wise elimination of non-significant co-factors was used, and the most complex non-significant terms were always removed first. When only significant terms remained, the P value for the independent variable (group x time) was obtained, which was corrected for the effects of any remaining significant terms.

9.3.3 Initial comparison of strength only (SO) vs strength and Tai Chi (STC) groups

There were no differences between the two exercise groups for all but 7 variables ($P<0.05$), which were the acceleration power on the step up by the less steady leg at 4-8Hz, isometric strength in the weak hamstrings at 80° and the strong hamstrings at 90°, RF CSA on both legs and isometric asymmetry in the hamstrings at 80° and 50°. The SO group showed greater increases in all these variables from pre to post-test except for asymmetry and steadiness variables, where a greater decrease was observed. That is, the SO group showed significantly better improvements for all seven variables. For brevity, Table 9.2 only shows results for the 7 variables with a significant intervention group x time interaction.

Variable	SO		STC		P
	pre	post	pre	post	
Acceleration power during the Step up (less steady leg) in 4-8Hz band ($rad.sec^{-2})^{-2}$	108 (21)	107 (12)	145 (18)	83 (11)	0.041
Isometric strength (N) weak hams 80°*	83 (9)	135 (10)	108 (9)	134 (9)	0.006
Isometric strength (N) strong hams 90°	97 (9)	134 (10)	111 (9)	126 (8)	0.042
RF CSA (cm^2) large leg*	3.5 (0.4)	6.2 (0.5)	4.2 (0.4)	5.3 (0.5)	0.002
RF CSA (cm^2)small leg*	2.9 (0.5)	6.6 (0.6)	3.5 (0.4)	5.3 (0.5)	0.002
Symmetry hams 80° (%)	26 (3)	16 (4)	12 (3)	15 (3)	0.025
Symmetry hams 50° * (%)	28 (4)	6.5 (3)	18 (4)	16 (3)	0.009

Table 9.2. Comparison of significant ($P<0.05$) pre to post-test changes between the two intervention groups for variables found to differ between the two intervention groups. * corrected for cofactors (see Table 9.2). hams = hamstrings.

During this preliminary analysis, cofactors interacted with some of the steadiness, strength, power and asymmetry variables (P<0.05) and all such interactions (including those for variables where the variables did not show significant group x time interactions) are shown in Table 9.3. Variables not shown did not interact with any cofactors. Group x time comparisons were corrected for these interactions.

Variable	Interacting factor	Direction of interaction with variable
Step up 1-4Hz less steady	Faller status x time	↓ more in fallers
Step up PF/DF less steady	Sex x time	↑ more in men
Step down PF/DF less steady	Sex x time	↑ more in men
Isometric hams 50° weak leg	Sex x time	↑ more in men
Isometric hams 80° weak leg	Sex x time	↑ more in men
Isometric hams 90° strong leg	Sex x time	↑ more in men
eccentric quads 150°.sec^{-1} weak leg	Faller status x time	↑ more in non-fallers
Symmetry quads 50°	Faller status x time	↓ more in non-fallers
Symmetry hams 50°	Sex x time	↓ more in men
Symmetry hams 60°	Sex x time	↓ more in men
RF CSA larger leg	Sex x time	↑ more in men
RF CSA smaller leg	Sex x time	↑ more in men

Table 9.3 Cofactors interacting with steadiness, strength, CSA and asymmetry variables. PF/DF=plantarflexion to dorsiflexion activation ratio.

9.3.4 Non-pooled analysis

The 7 variables with a significant intervention group x time interaction were then analysed between three groups (SO *vs* STC *vs* control). There was a significant group x time interaction for weak isometric hamstrings at 80°, RF CSA bilaterally and hamstring asymmetry at 50°. After post hoc testing, the only group x time interaction involving a difference relative to the control group was for the smaller leg RF CSA, where the SO group had a greater increase in CSA than the control group (P<0.05). As expected from the preliminary analysis, there were significant post-hoc group x time interactions between the SO and STC groups for isometric hamstring strength at 80° in the weaker leg (P<0.05), RF CSA bilaterally (P<0.01) and hamstring asymmetry at 50° (P<0.05) (Table 9.4).

	SO		STC		Control		
	pre	post	pre	post	pre	post	Anova
SU 4-8Hz less steady leg	100 (23)	106 (12)	156 (21)	84 (11)	131 (18)	91 (10)	0.072
Isometric hams 80° weak leg §	80 (9)	126 (10)	106 (9)	126 (9)	84 (8)	102 (8)	0.018
Isometric hams 90° strong leg	93 (9)	123 (10)	107 (9)	117 (10)	104 (8)	105 (9)	0.052
RF CSA larger leg §	3.4 (0.4)	5.4 (0.5)	4.1 (0.4)	4.7 (0.5)	4.8 (0.5)	5.3 (0.6)	0.013
RF CSA smaller leg +§	3.1 (0.5)	6.5 (0.5)	3.6 (0.4)	5.3 (0.5)	4.1 (0.5)	5.6 (0.6)	0.002
Symmetry hams 80°	26 (3)	16 (4)	12 (3)	15 (3)	18 (3)	16 (3)	0.133
Symmetry hams 50° §	28 (4)	8 (3)	17 (4)	17 (3)	18 (4)	11 (3)	0.019

Table 9.4 Pre and post-test values in the control and intervention groups (for variables found to differ between the two intervention groups). + = significant post hoc group x time interaction between the strength only and control group (P<0.05). § = significant post hoc group x time interaction between the strength only and Tai Chi and strength groups (P<0.05). SU=step up. (NB: Some of the mean values for the intervention groups are different to those in table 9.3 as they are corrected for different/no cofactors. The cofactors that were significantly different depended on the groups considered.)

Interactions with cofactors were as outlined in Table 9.5, and the group x time comparisons were corrected for these.

Variable	Interacting factor	Direction of interaction with variable
Step up 4-8Hz less steady leg	Faller status x time	↓ more in fallers
RF CSA small	Sex x time	↑ more in men
Symmetry hams 50°	Sex x time	↓ more in men

Table 9.5 Cofactors interacting with variables in the non-pooled analysis. Sym=Symmetry

9.3.5 Pooled analysis

For the vast majority of variables the intervention groups were pooled, and cofactors interacted with some of the steadiness, strength, power and asymmetry variables (P<0.05) as shown in Tables 9.6 and 9.7. Variables not shown did not interact with any cofactors. Group (pooled exercise *vs* control) x time effects were corrected for these interactions.

Variable:	Interacting factor(s)	Direction of interaction with variable
Stand steadier leg	Group x faller status x time	Exercise fallers ↑ most
Step up steadier leg 1-4Hz	faller status x time	Fallers ↓ most
Step up steadier leg 18-32Hz	faller status x time	Non-fallers ↓ most
Step up less steady leg 1-4Hz	faller status x time	Fallers ↓ most
Step down less steady leg 1-4Hz	faller status x time	Fallers ↓ most
Step down steadier leg 1-4Hz	faller status x time	Fallers ↓ most
Step down less steady leg 4-8Hz	Group x faller status x time	Non faller control ↑ most
Sit less steady leg 1-4Hz	faller status x time	Non-fallers ↓ most
25% MVC steadier leg	Sex x time	Women ↓ most
25% MVC less steady leg	faller status x time	Fallers ↓ most
10% MVC less steady leg	faller status x time	Fallers ↓ most
Step down PF/DF steadier leg	Sex x time	Women ↓ most

Table 9.6 Cofactors interacting with steadiness and co-activation variables in the pooled analysis. PF/DF = plantarflexion to dorsiflexion activation ratio.

Variable	Interacting factor(s)	Direction of interaction with variable
Quads 60° strong leg	Sex x time	Men ↓ most
Quads 50° strong leg	Sex x time	Men ↓ most
Quads 50° weak leg	faller status x time	Fallers ↓ most
Quads C 150°.sec^{-1} weak leg	Sex x time	Women ↓ most
Sym quads 90°	Sex x time	Women ↓ most
Sym quads 80°	Group x faller status x time	Non faller control ↑ most
Sym quads 50°	faller status x time	Non-fallers ↓ most
Sym hams 60°	Sex x time	Men ↓ most
Sym quads concentric 50°.sec^{-1}	Sex x time	Men ↓ most
Sym quads concentric 150°.sec^{-1}	Sex x time	Men ↓ most
Sym power	Sex x time	Women ↓ most

Table 9.7 Cofactors interacting with strength and asymmetry variables in the pooled analysis. Sym=symmetry

9.3.5.i Steadiness

There were no group x time interactions for any isometric, anisometric or functional steadiness variable, indicating that the changes from pre to post-test did not differ between pooled exercise and control groups. Averaging across variables within each steadiness variable type did not reveal any group x time interactions (See Appendix 7, Tables A7.1-A7-3).

9.3.5.ii Fourier analysis of the step up, step down, stand and sit tasks

There were no group x time interactions for any acceleration power variable in the 4 functional tasks, indicating that the changes from pre to post-test did not differ between pooled exercise and control groups. When each band was averaged across tasks and legs there were also no significant group x time interactions (See Appendix 7, Tables A7.8).

9.3.5.iii Co-activation

There were significant group x time interactions in the step down on the less steady leg and the stand on the steadier leg, where the plantarflexor/dorsiflexion (PF/DF) activation ratio showed a greater decrease from pre to post-test in the pooled exercise than control group. Averaging of values was not carried out because incomplete data sets led to very small groups after averaging (Tables 9.8-9.9).

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean(SE)	Mean(SE)		Mean(SE)	Mean(SE)
Stand H/Q steadiest +	10	1.24 (0.23)	0.51 (0.09)	7	1.19 (0.19)	1.55 (0.48)
Stand H/Q less steady	11	1.16 (0.16)	0.62 (0.15)	8	1.43 (0.41)	1.05 (0.30)
Sit H/Q steadiest	10	0.85 (0.12)	0.60 (0.10)	7	5.28 (4.02)	0.62 (0.14)
Sit H/Q less steady	12	1.01 (0.17)	0.61 (0.13)	8	0.74 (0.12)	0.99 (0.27)
Step up H/Q steadiest	10	1.05 (0.16)	0.64 (0.15)	9	1.09 (0.14)	0.64 (0.12)
Step up H/Q less steady	10	1.06 (0.16)	0.52 (0.10)	7	0.78 (0.11)	0.51 (0.14)
Step down H/Q steadiest	13	0.90 (0.15)	0.55 (0.09)	8	0.94 (0.16)	0.47 (0.13)
Step down H/Q less steady	9	1.23 (0.22)	0.48 (0.09)	9	1.16 (0.39)	0.60 (0.15)

Table 9.8 Pre and post-test measurements of co-activation across the knee joint during functional tasks in exercise and control groups. += significant group x time interaction (P<0.05). H/Q = hamstring:quadriceps activation ratio

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean(SE)	Mean(SE)		Mean(SE)	Mean(SE)
Stand PF/DF steadiest	11	0.84 (0.20)	0.41 (0.06)	7	1.07 (0.22)	0.53 (0.19)
Stand PF/DF less steady	10	0.93 (0.15)	0.53 (0.17)	8	1.44 (0.37)	0.51 (0.16)
Sit PF/DF steadiest	10	1.01 (0.35)	0.36 (0.05)	7	0.62 (0.19)	0.42 (0.11)
Sit PF/DF less steady	12	1.05 (0.25)	0.34 (0.07)	7	0.80 (0.10)	0.25 (0.04)
Step up PF/DF steadiest	10	1.24 (0.24)	0.87 (0.39)	10	2.30 (1.48)	0.70 (0.15)
Step up PF/DF less steady	11	0.90 (0.16)	1.63 (0.26)	10	1.14 (0.27)	1.02 (0.20)
Step down PF/DF steadiest	9	0.42 (0.12)	0.86 (0.16)	8	0.78 (0.12)	0.60 (0.15)
Step down PF/DF less steady +	10	1.11 (0.24)	0.77 (0.26)	9	0.63 (0.08)	0.49 (0.09)

Table 9.9 Pre and post-test measurements of co-activation across the ankle joint during functional tasks in exercise and control groups. += significant group x time interaction (P<0.05). PF/DF = plantarflexion:dorsiflexion activation ratio.

A correlation of the percentage change in co-activation with the percentage change in steadiness showed that there was no relationship between the variables (table 9.10).

		STAND	SIT	Step up	Step down
PF/DF Steadiest	R	0.12	-0.07	-0.08	0.35
	P	0.65	0.80	0.74	0.17
	N	17	17	18	17
PF/DF Less steady	R	-0.05	-0.12	0.13	0.47
	P	0.85	0.68	0.61	0.08
	N	15	15	17	15
H/Q Steadiest	R	-0.10	-0.15	-0.09	-0.45
	P	0.71	0.57	0.74	0.09
	N	17	17	17	15
H/Q Less steady	R	-0.16	0.41	0.14	0.38
	P	0.54	0.10	0.61	0.12
	N	16	17	15	18

Table 9.10 Correlation between steadiness during functional manoeuvres (columns) and co-activation of plantarflexors and dorsiflexors (PF/DF) and co-activation of hamstrings and quadriceps (H/Q). There were no significant associations. The steadiness and co-activation values used are from the same leg.

9.3.5.iv Asymmetry of steadiness

There were no group x time interactions for steadiness asymmetry. Averaging of values also did not reveal an overall group x time interaction (Appendix 7, Table A7.9).

9.3.5.v Isometric strength

There were significant group x time interactions for quadriceps strength at 60, 80 and 90° flexion in both legs, and in hamstring strength at 30 and 50° in the weak leg (Figs. 9.1 - 9.4). In all these cases, the pooled exercise group showed greater pre- to post- increases. When values were averaged across muscles, angles and sides there was a significant group x time interaction (pooled exercise (ex) pre: 200±9 N, ex post 235±9 N, control (cont) pre 186±14N, cont post 197± 14 N, P<0.01).

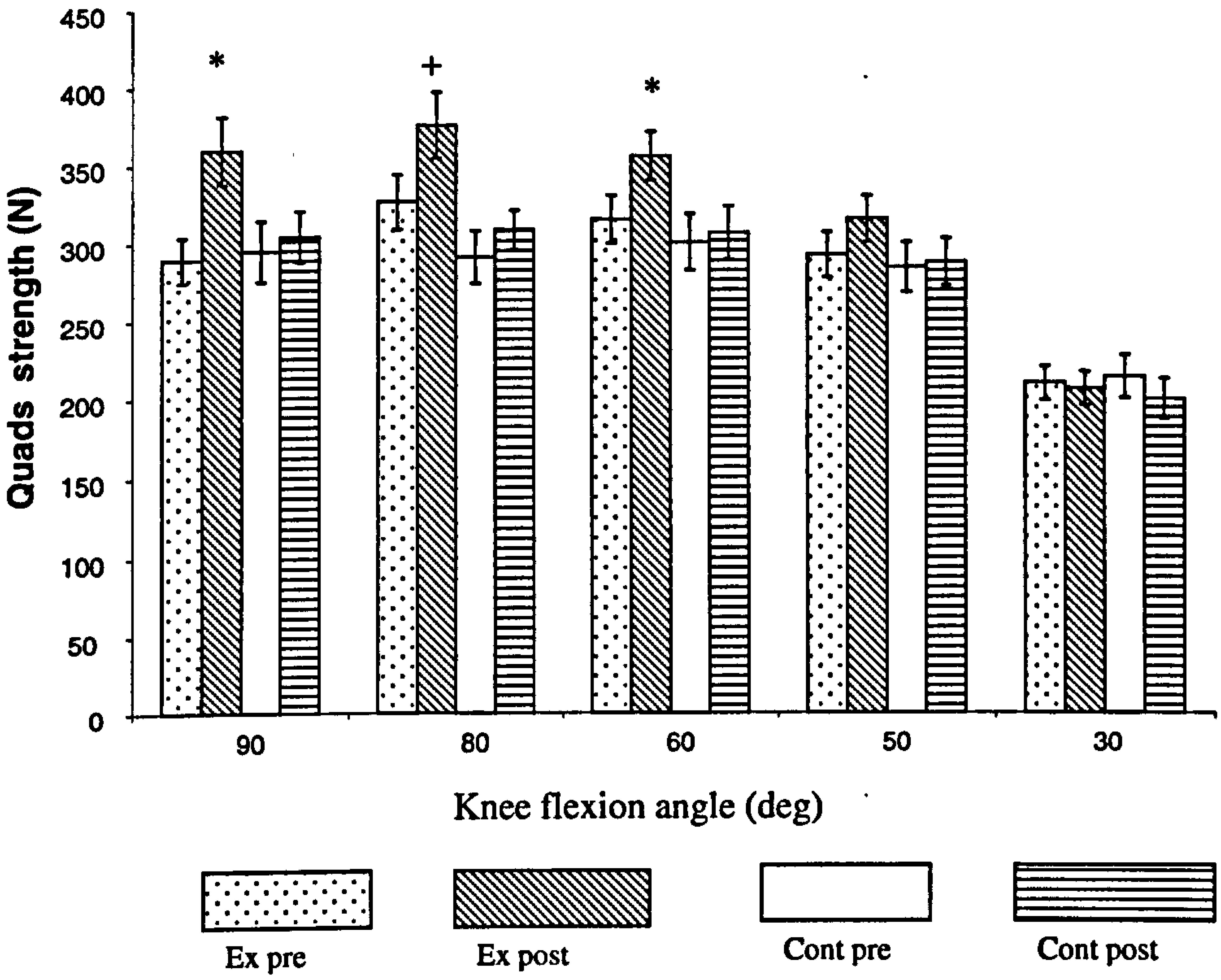


Fig. 9.1 Pre and post-test measurements of isometric quadriceps strength in the exercise and control groups in the stronger leg. * significant group x time interaction (P<0.01), + significant group x time interaction (P<0.05). Ex = exercise group, Cont =control group, pre = pre-test, post = post-test, N = Newton

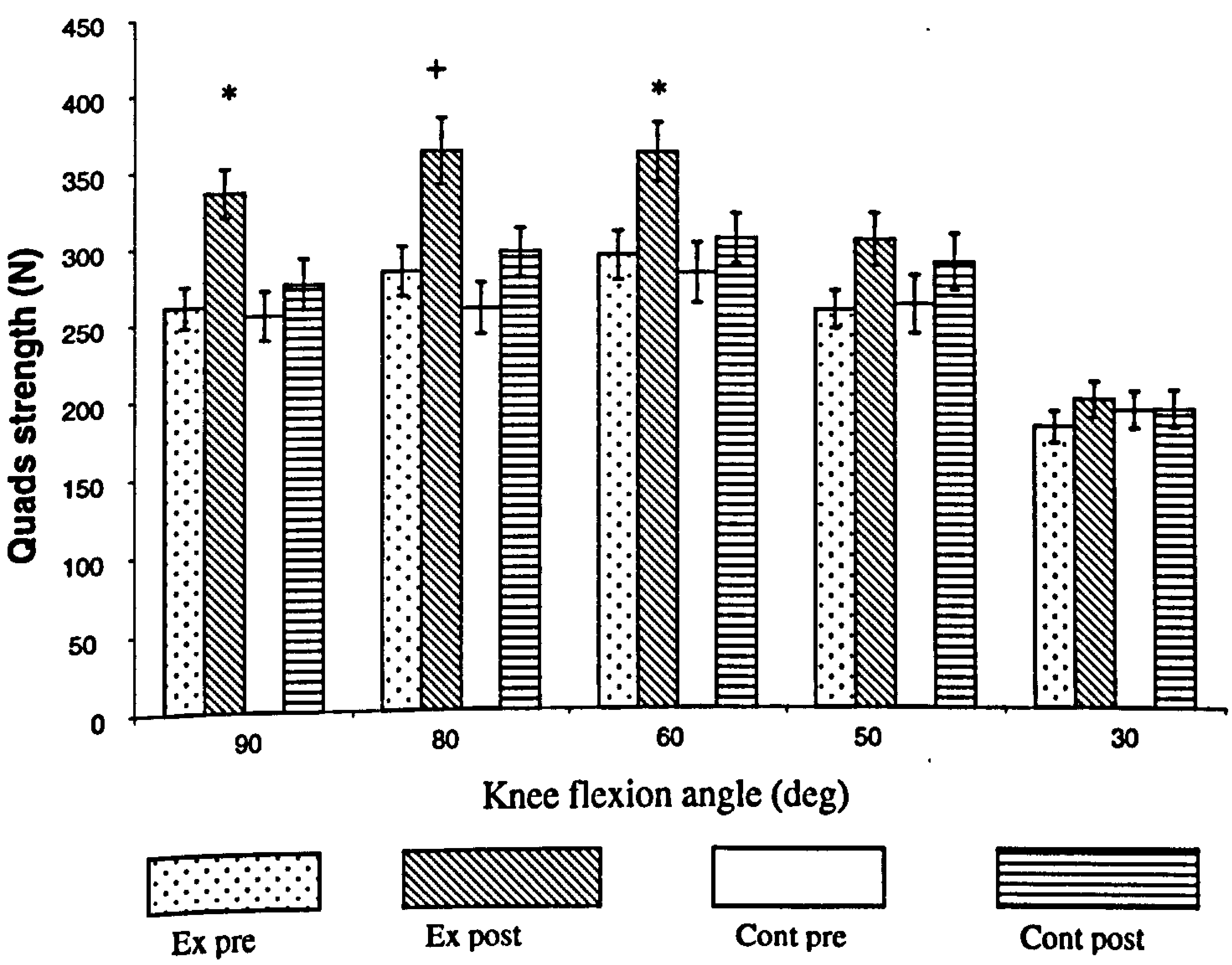


Fig. 9.2 Pre and post-test measurements of isometric quadriceps strength in the exercise and control groups in the weaker leg. * = significant group x time interaction (P<0.01) + = significant group x time interaction (P<0.05). Ex = exercise group, Cont=control group, pre = pre-test, post = post-test, N = Newton

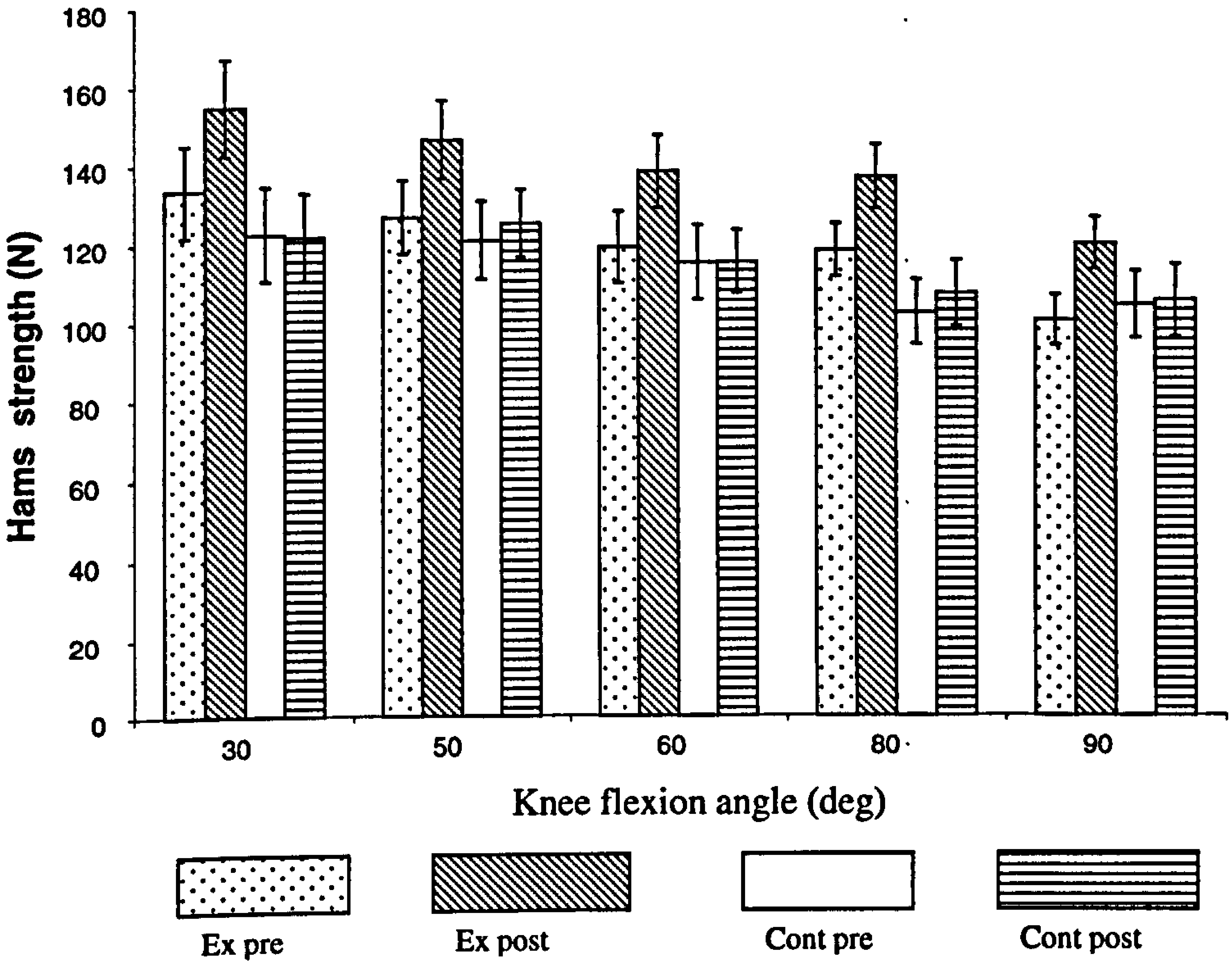


Fig. 9.3 Pre and post-test measurements of isometric hamstring strength in the exercise and control groups in the stronger leg. There were no significant group x time interactions. Ex= exercise group, Cont=control group, pre = pre-test, post = post-test, N = Newton

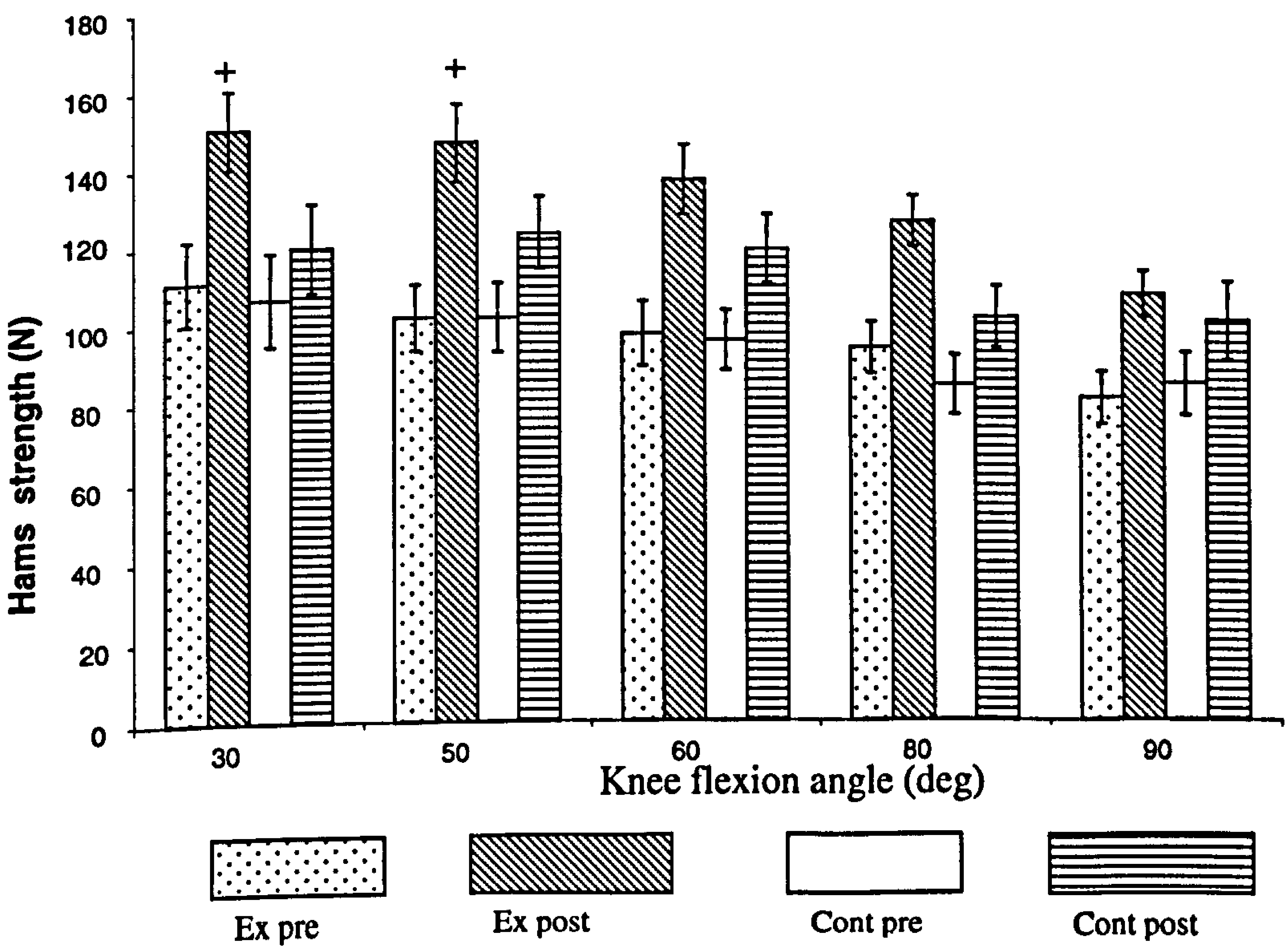


Fig. 9.4 Pre and post-test measurements of isometric hamstring strength in the exercise and control groups in the stronger leg. + = significant group x time interaction ($P<0.05$). Ex= exercise group, Cont = control group, pre = pre-test, post = post-test, N = Newton

9.3.5.vi **Isokinetic strength**

There were significant group x time interactions for weak quadriceps eccentric strength at 50°.sec^{-1}, strong hamstring eccentric strength at 150°.sec^{-1} and weak hamstring concentric strength at 50°.sec^{-1} (Figs. 9.5 - 9.6). When values were averaged across muscles, angles and sides there was a significant group x time interaction for the concentric contractions (ex pre: 145$\pm$10 N, ex post 176$\pm$11 N, cont pre 146$\pm$14 N, cont post 149$\pm$15 N, P<0.01) and a significant group x time interaction for eccentric contractions (ex pre: 268$\pm$ 15 N, ex post 306$\pm$15 N, cont pre 255$\pm$20 N, cont post 255$\pm$20 N, P<0.01). In all significant cases, the pooled exercise group showed greater increases.

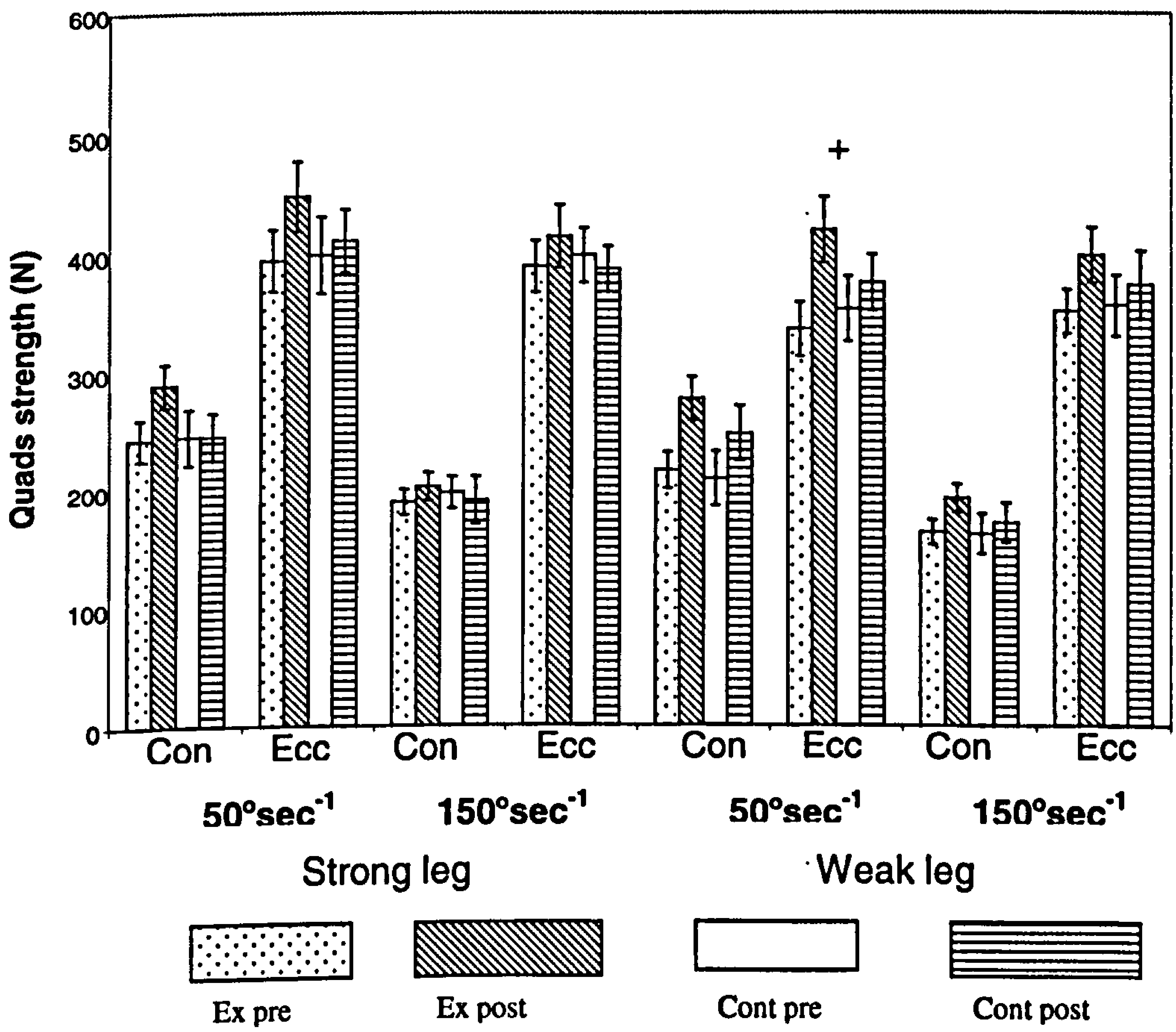


Fig. 9.5 Pre and post-test measurements of isokinetic quadriceps strength in the exercise and control groups. + = significant group x time interaction (P<0.05). S = stronger leg, We = weaker leg. Ex = exercise group, Cont = control group, pre = pre-test, post=post-test , N = Newton

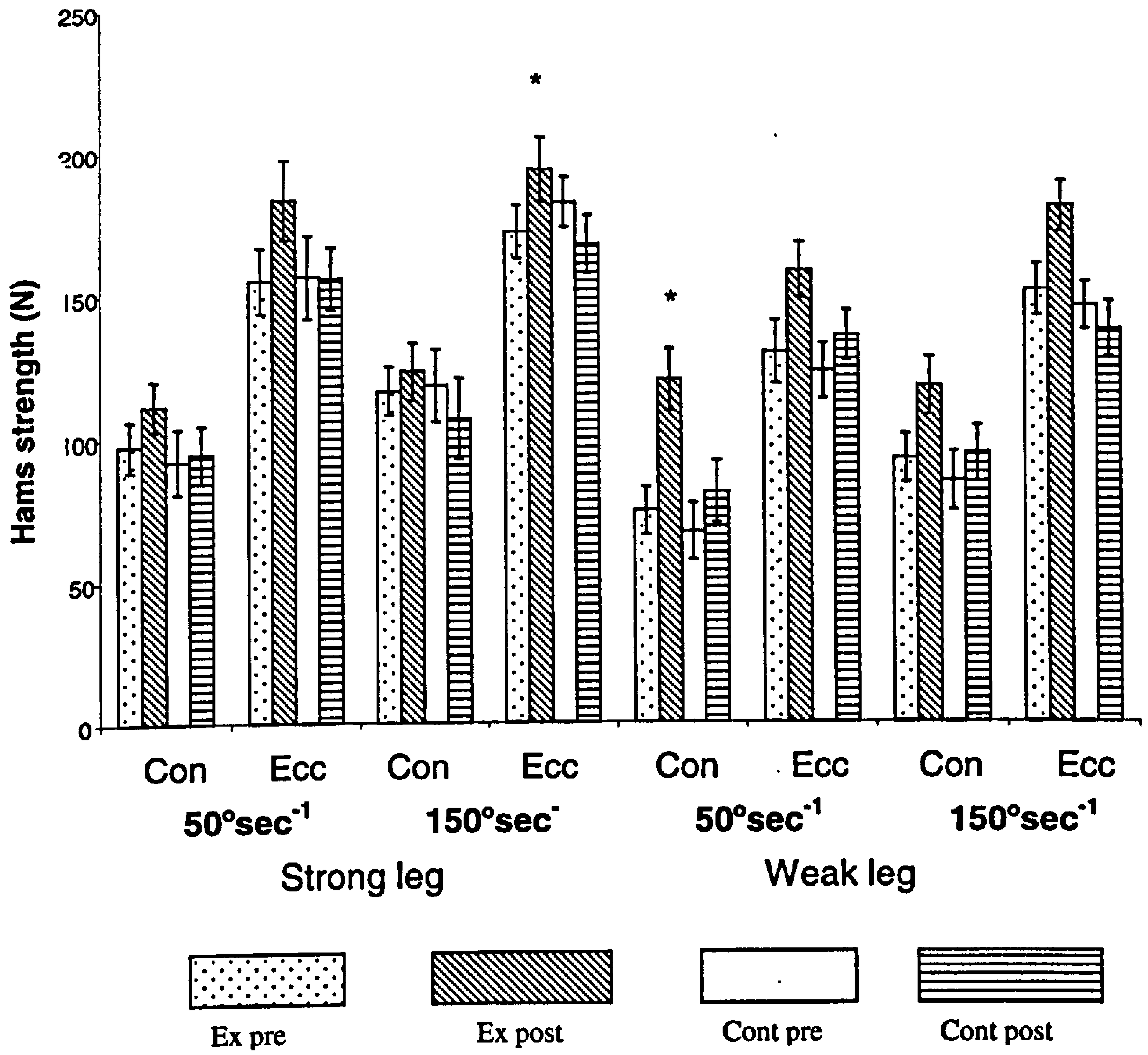


Fig. 9.6 Pre and post-test measurements of isokinetic quadriceps strength in the exercise and control groups. * = significant group x time interaction (P<0.01). S = stronger leg, We = weaker leg. Ex = exercise group, Cont = control group, pre = pre-test, post = post-test, N = Newton

9.3.5.vii Leg extension power

There were significant group x time interactions for leg extension power on both legs (Fig. 9.7). Pre to post increases were greater for the pooled exercise group.

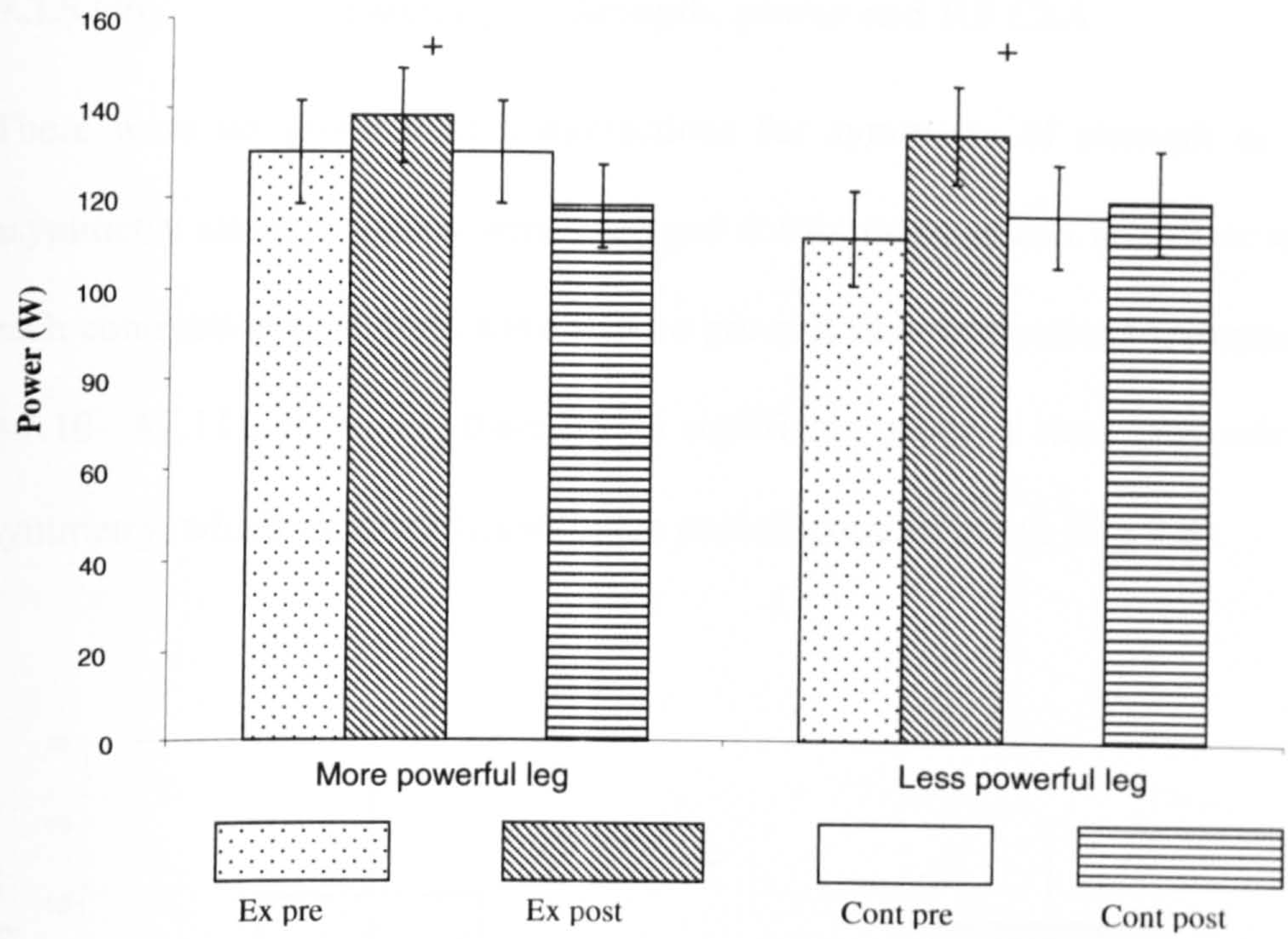


Fig. 9.7 Pre and post-test measurements of power in the exercise and control groups in the stronger and weaker legs. + = significant group x time interaction (P<0.05). Ex = exercise group, Cont = control group, pre = pre-test, post = post-test, W=Watts

Calculation of mean percentage increases for the pooled exercise and control groups for quadriceps strength at 90° and leg extensor power showed that when control changes were subtracted from pooled exercise group changes, strength changes were numerically greater than power changes (table 9.11).

Variable	Exercise %↑	Control %↑	Exercise %↑- Control %↑
Quads 90° strong leg	24.2	3.1	21.1
Quads 90° weaker leg	28.3	7.8	20.5
Power more powerful leg	6.1	-10.0	16.1
Power less powerful leg	20.9	4.3	16.6

Table 9.11. Percentage increases in strength and power in the exercise and control groups. After allowing for the control group, power increases were numerically lower than strength increases.

9.3.5.viii Symmetry of strength, power and RF CSA

There were no group x time interactions for symmetry of strength or CSA. When asymmetry strength values were averaged across muscles and angles or speeds within each contraction type there were also no group x time interactions (Appendix 7, Tables A7.10- A7.11). However, there was a significant group x time interaction for power symmetry, which improved more in the pooled exercise group (Fig 9.8).

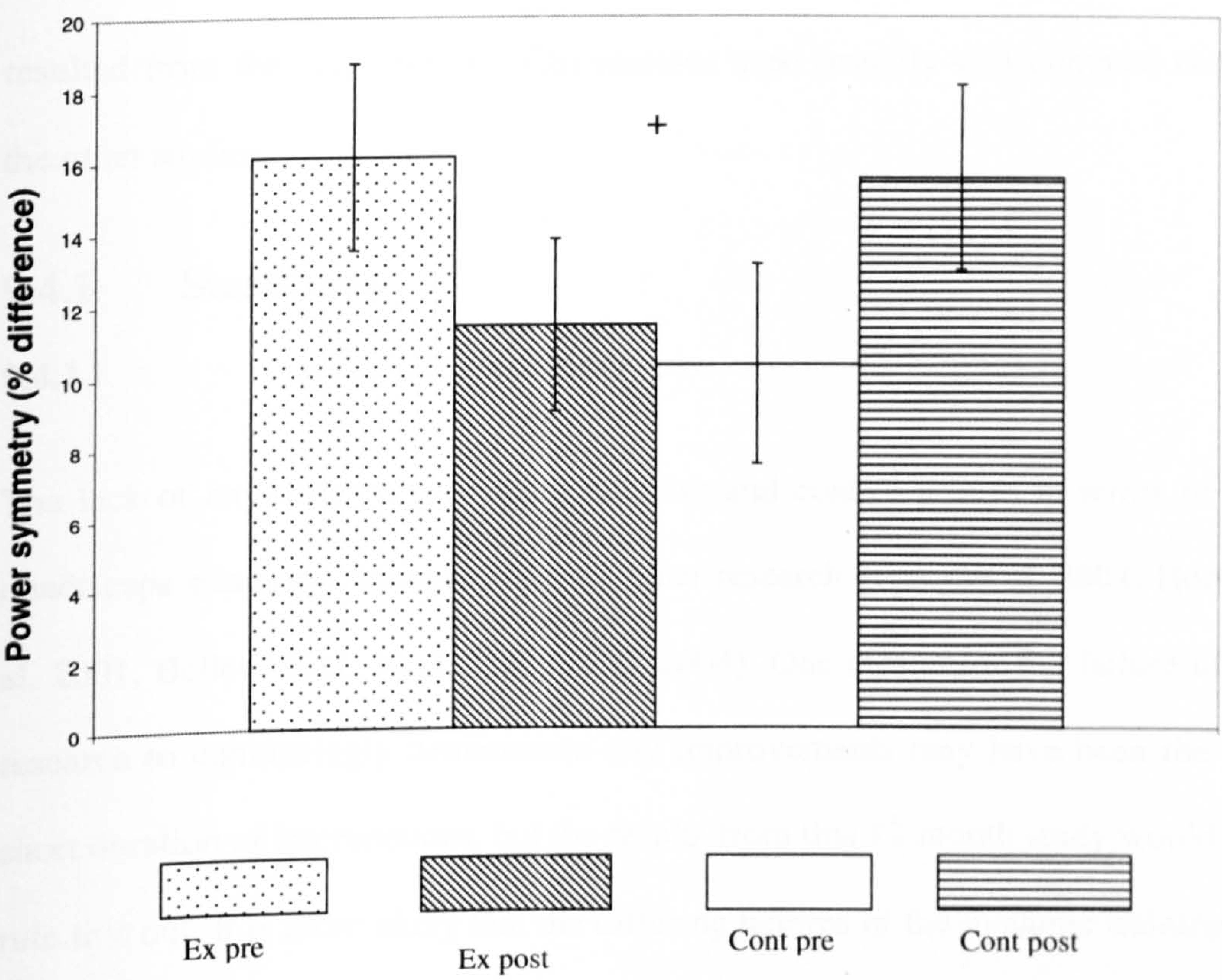


Fig. 9.8 Pre and post-test measurements of power symmetry in the exercise (n=30) and control groups (n=25). + = significant group x time interaction (P<0.05).

9.4 Discussion

As the addition of Tai Chi only influenced the outcome of one variable (RF CSA in the smaller leg) in relation to the control group, and that Tai Chi as used in this study was therefore generally inert, the following discussion will assume that the intervention was strength training alone, unless stated otherwise. This study thus contradicts other research that has found Tai Chi can improve steadiness (Christou et al. 2003b, Yan et al. 1999) and strength (Christou et al. 2003b, Lan et al. 1998). This may have partially resulted from the very brief Tai Chi sessions used (usually <10 minutes) compared to the other studies.

9.4.1 Steadiness

9.4.1.i Isometric steadiness

The lack of any difference between exercise and control groups in terms of isometric quadriceps steadiness changes supports other research (Tracy et al. 2001, Hortobagyi et al. 2001, Bellew et al. 2002, Tracy et al. 2004). One reason for the failure of previous research to convincingly demonstrate any improvements may have been the relatively short duration of interventions, but the results from this 12 month study would appear to rule this out. It is more likely that the differing natures of the dynamic training stimulus and the static steadiness test contributed to the lack of an effect. This, and the fact that the intervention improved isometric quadriceps strength, supports the concept that the training-related improvements seen in anisometric steadiness (Hortobgayi et al. 2001, Tracy et al. 2004) are more likely to be due to movement practice than an increase in strength (Laidlaw et al. 1999, Christou et al. 2003b, Tracy et al. 2004).

It should be noted that the test-retest reliability CoV of around 15-23% for the three isometric contraction intensities (Appendix 6) was generally more than the pre to post mean change within each group, and much greater than the difference between group changes. This means that any real effects may have been masked by measurement noise. However, there was no noticeable trend towards any group difference and so this is unlikely to be the sole reason for the results.

9.4.1.ii Anisometric steadiness

The lack of any training effects on concentric or eccentric steadiness in this study contrasts with other findings. Hortobagyi et al. (2001) and Tracy et al. (2004) demonstrated improvements in both contraction types in the quadriceps, whilst Laidlaw et al. (1999) demonstrated similar effects in hand muscles.

One explanation for the differing results in this study is a Type II error due to low subject numbers, which is confirmed by the observed power being only 0.07 for the group x time interaction for the concentric 1kg variable. Low numbers of subjects for this test resulted from problems with the equipment during the middle part of the pre-testing stage, and further studies with greater numbers are required.

9.4.1.iii Functional steadiness

This is the first time that the effects of exercise on functional steadiness have been investigated. The lack of any group differences in the pre- to post-test changes of the SD of knee angular acceleration, or spectrally analysed acceleration power, suggests that functional steadiness is not improved by the training used in this study. Again, limited test-retest reliability is unlikely to be solely responsible for the overall conclusion, as comparison of the exercise and control group mean changes do not indicate any trends.

A plausible reason for the lack of findings may be the differences between the training actions and functional task actions. Further studies using more functional exercises might lead to different results. Another potentially important factor was that the speed of movement was not controlled. Subjects were allowed to select their preferred speed, and this might have meant that any power or strength gains from the training led to increased speed during the functional steadiness tasks in the post-test. This might have increased acceleration fluctuations (Christou et al. 2003a), potentially cancelling out any steadiness improvements that might have been observed if the speeds had been kept constant.

9.4.1.iv Effects of training-related co-activation changes on steadiness

The greater decreases in co-activation in the exercise group during the functional tasks suggest that the training was responsible for a reduction in co-activation. These changes, however, did not affect functional steadiness, which was confirmed by a lack of correlation between percentage changes in functional steadiness and co-activation. This agrees with the baseline finding of no significant relationship between co-activation levels and functional steadiness in the older subjects (Chapter 7).

This is in contrast to other findings in the FDI (Laidlaw et al. 1999) and ankle (Patten and Kamen 2000) where changes in steadiness and co-activation were observed to co-exist, although the strength of any relationship was not assessed through a correlation analysis.

9.4.1.v Asymmetry of steadiness

Training did not influence steadiness asymmetry, which is unsurprising since training did not influence steadiness. Steadiness asymmetry appears to increase with age, but greater asymmetry does not appear to be related to falling. Further work is required to evaluate its significance.

9.4.1.vi Influence of faller status on steadiness changes after training.

Despite the lack of any training effects on steadiness, analysis of the faller status x time and group x faller status x time interactions within the model is revealing. For the variables where faller status x time interacted with the independent variable (Table 9.5), fallers generally showed a greater improvement in steadiness, but as group x time interactions were not significant, this was across both the exercise and control groups, and so does not demonstrate that fallers respond better to training.

There were also two variables with significant group x faller status x time interactions. In the step down manoeuvre, acceleration power increases in the 4-8Hz band were significantly higher in the non-faller controls. This suggests that the non-faller exercise group had a relative improvement in steadiness compared to the non-faller controls, which was not evident amongst the fallers in the exercise group. In turn this suggested that only non-fallers responded to training. Similarly, during the standing manoeuvre on the steadier leg, the fallers in the exercise group showed a significantly higher increase in acceleration fluctuations than the other groups. This indicated that training worsened functional steadiness during standing in fallers. Together, these findings suggested that only non-fallers showed a tendency to improve steadiness with training.

This is a surprising result. If, as other studies have suggested, training can reduce force fluctuations, fallers might be expected to improve more readily since they may have poorer steadiness to start with (Chapter 6). Tracy et al. (2004) showed a strong positive association between the absolute training related improvement in steadiness and the level of unsteadiness at baseline, which supports this assertion. These isolated findings must therefore be treated with caution, and further work is required.

9.4.2 Strength, power and RF CSA

9.4.2.i Effects of training on isometric strength.

Strength training increased knee extensor and flexor isometric strength in the elderly relative to a control group, as shown by the averaged results across muscles, angles and legs. However, the degree of this adaptation varied between muscles and joint angles.

In line with previous research (Sipila et al. 1996, Skelton et al. 1995, Hakkinen et al. 2001, Lord et al. 1995, Meuleman et al. 2000, Hortobagyi et al. 2001, Seynnes et al. 2004, De Vreede et al. 2005) isometric quadriceps strength increased with training. These improvements only occurred between 90-60° knee flexion, and not inner quadriceps ranges, which may relate to the range used during knee extensor training. Test-retest reliability may also be poorer at shorter muscle lengths (unpublished observations, Appendix 1) which might also explain the lack of detected differences.

The isometric strength improvement in the hamstrings was more limited, with training increases only observed in the weaker leg at 30 and 50° knee flexion. In contrast, Lord et al. (1995) observed small but significant improvements at 90° knee flexion after one year of twice weekly training, although details of the actual hamstring exercise were not given. One reason for the lack of improvement at shorter lengths may be a difference in position between training and testing, with the former done in prone with hip extension. This means the hamstrings were unlikely to have been trained at the shorter muscle lengths. Another possibility was the poorer test-retest reliability at shorter hamstring lengths (Appendix 1). Overall, these results suggest that hamstring strengthening is possible in the elderly and since hamstring isometric strength appears to be particularly associated with falls (Chapter 4), hamstring training may be of value in falls prevention.

9.4.2.ii Effects of training on concentric strength

Concentric strength in the quadriceps and hamstrings was also increased by strength training, as shown by the averaged results across muscles, speeds and legs. This concurs with previous work (O'Neill et al. 2000, Hortobagyi et al. 2001, Hakkinen et al. 2001, Cress et al. 1999, Capodiaglo et al. 2005). Again, however, the amount of change depended on the muscle and speed.

When variables were analysed individually, training only led to improvement in the hamstrings at $50°.sec^{-1}$. Test-retest reliability is unlikely to explain the lack of other differences as reliability was high for measures not showing a group difference (Appendix 1). Low numbers in the control group may partially explain results, with an observed power of only 0.33 for the slow concentric quadriceps variable for group x time interactions.

9.4.2.iii Effects of training on eccentric strength

The greater improvements in the intervention group for two eccentric variables and the average of all eccentric variables, agreed with the findings of Hortobagyi et al. (2001), but differed from those of Reeves et al. (2005). Reeves et al. (2005) suggested that the lack of eccentric improvements was partly due to eccentric loading being insufficient, as load was prescribed according to concentric strength, which is lower. This effect may have been further heightened by the relative preservation of eccentric strength (Reeves et al. 2005). This effect may explain the lack of significant improvement in the other 6 eccentric contraction conditions when each was considered separately. Low control subject numbers may also be responsible, with an observed power of only 0.35 for the fast eccentric quadriceps variable for group x time.

Despite the relative preservation of eccentric strength with age compared to concentric strength, eccentric strength does decrease with age (Chapter 3), and reductions in eccentric strength have also been noted in fallers (Chapter 4). Hence more effective eccentric training than is possible with standard isotonic training may be important. Use of isokinetic machines for training may be one strategy to achieve this, as they can deliver maximal resistance during eccentric contractions. Further studies should investigate the effects of this form of training on eccentric strength in older people.

9.4.2.iv Effects of training on RF CSA

Change in RF CSA in the smaller leg was greater in the strength only group than the controls and the strength and Tai Chi group. This was the only one of the seven outcome variables differing between intervention groups that showed a difference relative to the controls. This negative effect of adding Tai Chi to strength training was unexpected. Since it is known that Tai Chi did not lead to reduced time performing strength training, this either implies that Tai Chi and strength training were antagonistic or that the differences in RF CSA change between intervention groups were due to other unrelated factors. Antagonistic effects are seen when mechanisms of action are mutually incompatible (such as bacteriostatic and bactericidal antibiotics) but there is no evidence to suggest that the supposed mechanisms of Tai Chi's actions (improvement in muscle co-ordination (Christou et al. 2003b)) would be antagonistic to strength training mechanisms. Thus it is likely that these intervention group differences are independent of the intervention type, and it is possible that other factors such as differing instructors may have had an effect, as Tai Chi was only conducted by a single instructor.

This result is in line with previous research reporting that strength training can increase CSA in older people (Frontera et al. 1988, Fiatarone 1990, Roman et al. 1993, Pyka et al. 1994, Fiatarone 1994, Welle et al. 1996, Harridge et al. 1999, Tracy et al. 1999, Ferri et al. 2003), indicating that strength increases are accompanied by morphological changes. The lack of any difference between groups in the larger leg is therefore surprising, but this inconsistency may relate to the use of ultrasonography to measure CSA, which is the only measurement modality in the literature that has failed to show an increase in CSA with training (Siplila and Suominen. 1996). It has been shown that ultrasonography measurements of muscle CSA agree closely with MRI measurements (Walton et al. 1997, Reeves et al. 2004a), but these studies only considered the total CSA of tissue within the muscle, and did not attempt to compare measurements when non-contractile areas were excluded. As ultrasonography does not normally allow estimation of fatty infiltrates within a muscle as easily as MRI (Klein et al. 2001), any growth of muscle may not be detected as easily, as this may be offset by likely simultaneous decreases in non-contractile intramuscular tissues (Sipila and Suominen 1996). Another possibility is that the use of ACSA led to an apparent reduction of measured gains. Any hypertrophy might incease pennation angle (Kawakami et al. 1993) which tends to lead to ACSA being a greater underestimation of PCSA than otherwise. Finally, there may be variation in the amount of hypertrophy along different sections of the muscle belly (Narici et al. 1996), and it is possible that the measurement in one section failed to identify an increase. This may have been accentuated by the use of a relatively small quadriceps muscle to assess hypertrophy, which may have led to greater measurement errors.

Evaluation of training effects on F/CSA was not possible due to very low numbers of control subjects with both RF CSA and isometric strength measures for both pre- and post-tests.

9.4.2.v Effects of training on power

This is the fourth controlled study to show power increases from standard low velocity 8-10RM resistance training. Overall therefore this study's evidence tips the balance in favour of slow velocity strength training having a real effect on power. Given that decreased normalised power has been shown to be associated with falls in this and other work (Skelton et al. 2002) this suggests that standard resistance training may be able to reduce falls risk through increasing power. However, as argued in Chapter 4, it may not be power but one of its constituents – speed - that is the more important factor in reducing falls risk. To evaluate the effectiveness of this power increase in reducing falls it is therefore necessary to assess any changes in speed.

Although speed increases might be thought unlikely with slow velocity training, especially since strength training may lead to a conversion from IIb to IIa fibres in both young (Adams et al. 1993, Andersson et al. 1994) and older (Hakkinen et al 1998c) subjects, it has been shown that speed increases may occur in slow velocity strength training due to an increase in the stiffness of tendons (Reeves et al. 2003a). Stiffer tendons may enable a faster transfer of speed from the contractile fibres to the joint (Reeves et al. 2003a). Additionally, fascicle length increases with low velocity training in the elderly (Reeves et al. 2004b, 2004c, 2005) imply an increased number of sarcomeres and thus greater contraction speed (Jones and Round 1990). If such speed increases had occurred in this study, a greater increase in power than strength might be observed, as power increases would be compounded by the simultaneous increase in its constituents (a twofold increase in both speed and strength would lead to a fourfold increase in power). When allowing for concurrent control group increases, maximal isometric quadriceps strength (at 90° knee flexion) increased by 21.1% in the stronger leg and 20.5% in the weaker leg, but leg extension power increased by only 16.1 and

16.6% in the more and less powerful legs respectively. A statistical analysis to establish if the strength and power increases were truly different from each other was not possible when control group changes were accounted for. However, these results do confirm that power increases were not greater than strength increases, which is contrary to what might be expected if speed had increased along with strength.

Hence the improvement in power gained from slow velocity training may not automatically confer any greater protection from falling than the strength increases already described. As described in Chapter 4, although strength may assist in avoiding trips it may not be sufficient on its own to enable recovery. Despite the concerns of Newton and colleagues (2002) there are no reports in the literature of a greater rate of injury during high velocity training, and so the results from this study suggest that a combination of low and high velocity training may be the most effective form of training for older people.

9.4.2.vi Influence of faller status on strength, power and RF CSA

In all the strength and power variables showing a significant group x time interaction (that is, showing an improvement with training) there was no interaction with faller status x time, or group x faller status x time. This suggests that fallers and non-fallers do not differ in their ability to respond to strength or power training, which confirms previous findings (Robinson et al. 2004). This suggests that fallers should be able to reduce their strength and power deficits to levels where the risks of falling are reduced. This is supported by findings that suggest strength training can reduce falls (Campbell et al. 1997, Campbell et al 1999, Robertson et al. 2001, Mulrow et al. 1994).

9.4.2.vii Asymmetry changes

Strength training did not change asymmetry of strength or muscle CSA in the elderly. It is likely that training did not amend any asymmetries because both sides were equally trained in terms of both receiving the appropriate 80% 1RM load for each separate leg, which failed to remove the pre-existing difference between legs.

The finding that power asymmetry decreased with training is surprising, given the lack of change in asymmetry of strength. The implications of this are important. Power symmetry has been shown to be associated with frequent falls (Chapter 4, Skelton et al. 2002) and thus training may have the additional beneficial effect of reducing the risks from this source.

These results imply that future interventions should perhaps favour the less powerful leg to regain symmetry, if symmetry is seen as important in its own right. This would involve optimising training to the lesser leg and possibly holding back the progress of the better leg. However, as argued earlier (chapter 4) it may not be asymmetry *per se* that is a risk factor for falling, but the correlating level of reduced power in the less powerful leg. If so, the aim should be to train each leg as optimally as possible, and not to restrain progress of the more powerful leg. Further work on asymmetry as a risk factor is warranted.

9.4.3 Limitations of the intervention

Limitations of the intervention programme may provide additional reasons for the steadiness, strength and asymmetry results showing less response to exercise than previous evidence would suggest. First, the difference between the intervention and control groups in terms of exercise may have been less than expected. The activities of both groups in this study were not limited in any way. No activity diaries were utilised throughout the study, but it is known that several control subjects carried out their own informal training programmes during the course of the year. These included activities such as hill walking, sailing and even some resistance training. These were activities that the subjects participated in as part of their normal routine and it would not have been ethical or pragmatic to have limited them. The baseline Actigraph data suggest that both the exercise and control groups had similar background activity levels, which initially suggests that the net difference in terms of exercise between the groups was the intervention. However, these activity data were collected before the onset of the intervention, and it is possible that the exercise group may have reduced their pre-existing activity levels to accommodate the training programme, thus effectively reducing the net difference in exercise input between groups. In addition, the control group may have increased their activity levels as a result of their awareness that their physical performance would be tested again, or because their general awareness of their physical condition had increased. Whilst there are no data to support or refute these considerations, they should be borne in mind.

Another factor concerns the level of compliance with the intervention programme. Although no formal record was kept of attendance at the twice weekly classes by instructors (despite instructions to do so) some subjects only attended classes once a week, whilst others had absences of up to several weeks due to holidays or ill-health. Whilst the lack of documentation of attendance is a limitation to this study, the attendance may reflect normal adherence to exercise programmes in the community, and thus provides a realistic assessment of this form of intervention. However, the aggressiveness of the intervention was successfully maintained through regular contact with the exercise instructors, stressing the need to progress exercise load according to the 80% 1RM prescription, which in practice was observed by ensuring that the tenth repetition should be very difficult.

The intervention limitations, and those previously mentioned relating to testing methods, suggest that caution should be taken in interpreting these results, particularly those relating to anisometric and functional steadiness. Further work should use activity diaries and ensure that the training methods are related to the testing methods.

9.5 Conclusions

A 12 month period of twice-weekly lower limb strength training had no effect on isometric, anisometric or functional steadiness, but improved quadriceps and hamstring isometric strength. There were also concentric and eccentric improvements but these were not consistent. Lower limb power was increased, but this was due to strength and not velocity improvements, and so may not confer additional protection against falling. Although strength asymmetry did not change, power asymmetry improved. Skelton et al. (2002) and this study showed power asymmetry was related to frequent falling, and so this change may relate to a lower likelihood of falling. However, it may not be the asymmetry itself that causes the falling, but the associated loss of power in the worse leg.

Fallers and non-fallers did not show any differences in their response to training, indicating that fallers should be able to reverse their strength and power deficits as well as non-fallers.

10 Overall conclusions

This study has produced several findings that are either novel, or that supplement very limited previous data. These have principally been in the areas of symmetry and steadiness, and in specific differences between fallers and non-fallers.

Symmetry of concentric strength and steadiness may decline with age, but symmetry of isometric and eccentric strength, power output and CSA did not change, indicating that in general motor degeneration may be diffuse rather than focal. This concurs with the present understanding of strength and power output losses with ageing. Fallers were more asymmetrical in terms of eccentric and isometric strength, and frequent fallers were more asymmetrical in power output. Asymmetry may therefore be linked to falls in its own right or indirectly linked to falling through asymmetry being correlated to an especially weak or non-powerful leg. The data suggested that the latter is more likely, but further work is necessary to confirm this. Strength training improved power output symmetry, which may therefore partially explain the benefits of exercise for fallers.

Force steadiness in the quadriceps was shown to worsen with age during concentric contractions and during standing and stepping down. Reduced steadiness in stepping down was also associated with fallers, who also had worse isometric and eccentric steadiness. Descending stairs is associated with falls in community dwelling older people (Startzell et al. 2000) and so the worse steadiness during stepping down in fallers may indicate a reason for this. Better steadiness during low intensity anisometric contractions was also associated with better ambulatory function, and so maintenance of steadiness may be important in maintaining functional ability and quality of life in older people.

This study did not demonstrate any changes in steadiness in response to training in older fallers or non-fallers, but other studies have found quadriceps dynamic steadiness can be improved with training (Tracy et al. 2004, Hortobagyi et al. 2001), even on very low training intensities (Hortobagyi et al. 2001). This suggests that strength training for older subjects may be as important for steadiness improvement as for strength and power development.

Causes of age related decreases in lower limb steadiness in older people were not suggested by this study, but by exclusion of motor unit synchrony, antagonist co-activation and variability of motor unit firing, motor unit coherence appears to be a potential influence. Further work on the physiological causes of decreased steadiness may enable future research into physical or pharmacological approaches to improving steadiness.

Other novel findings about healthy fallers found in these studies included greater antagonist co-activation, which could contribute to falling by inducing delays, mechanical slowing and increased resistance to agonist movement. No deficits of muscle area in fallers relative to non-fallers suggest that the weakness associated with fallers is not solely due to sarcopenia, which is supported by reduced specific force in fallers. This was also the first study to evaluate strength at a variety of speeds and angles across different contraction types, confirming that reduced strength is associated with falling. It found that plantarflexor and hamstring strength are particularly important, and is the first study to show that falls are linked to reduced eccentric strength. It also indicated that reduced power output is linked to falls, although contrary to what has been assumed it also showed that fallers are not necessarily much slower than non-fallers. By exclusion, strength and power were shown to be possible causative factors. Fallers also appeared to respond to training as well as non-fallers, which suggests that training of fallers will be successful.

This study also contributed to knowledge in areas of controversy. It supported the suggestions that eccentric strength may be relatively spared by ageing, particularly at higher movement speeds. Rather surprisingly, it also suggested that strength during faster concentric contractions declines more slowly than strength during slower concentric contractions, but as this implies that speed may increase with age it is unlikely to be a true finding. This study also suggested that losses of muscle mass are more important than reductions in specific force in explaining sarcopenia, as specific force was unexpectedly found to rise with age.

As mentioned earlier, some of the differences noted between fallers and non-fallers were quite different to those seen between young and old. This is surprising, as in most cases the muscle function of the cohort of fallers was found to be consistent with greater physiological ageing, which is what would be expected given the fallers' apparent lack of pathology and the greater incidence of falling with age. This may mean that the fallers had unknown pathologies. It could also reflect the limitations of this study, or, most intriguingly, it could mean that healthy people do not only age at different rates but also in slightly different ways. Such different ageing pathways could contribute to some healthy people falling and some not. This is an area deserving of further study.

References

Aagaard P, Simonsen EB, Andersen JL, Magnusson SP, Halkjner-Kristensen J, Dyhr-Poulsen P 2000 "Neural inhibition during maximal eccentric and concentric quadriceps contractions: effects of resistance training. Journal of applied Physiology, vol 89, pp. 2249-2257

Adams, K. J., Swank, A. M., Berning, J. M., Sevene-Adams, P. G., Barnard, K. L., & Shimp-Bowerman, J. 2001, "Progressive strength training in sedentary, older African American women", *Medicine and Science in Sports and Exercise*, vol. 33, no. 9, pp. 1567-1576.

Adams GR. Hather BM. Baldwin KM. Dudley GA. 1993 "Skeletal muscle myosin heavy chain composition and resistance training". *Journal of Applied Physiology.* Vol. 74, pp. 911-915

Akima, H., Kano, Y., Enomoto, Y., Ishizu, M., Okada, M., Oishi, Y., Katsuta, S., & Kuno, S. Y. 2001, "Muscle function in 164 men and women aged 20-84 yr", *Medicine and Science in Sports and Exercise*, vol. 33, no. 2, pp. 220-226.

Alexander, N. B. 2001, "Falls and gait disturbances", *Clinical Geriatrics*, vol 9, no. 3, pp. 56-61.

Allen, G. M., Gandevia, S. C., & McKenzie, D. K. 1995, "Reliability of measurements of muscle strength and voluntary activation using twitch interpolation", *Muscle and Nerve*, vol. 18, no. 6, pp. 593-600.

Allum, J. H. J., Dietz, V. Freund H. J. 1978. " Neuronal Mechanisms Underlying Physiological Tremor". *Journal of Neurophysiology*, vol 41, pp. 557-571.

Allum, J. H. J., Carpenter, M. G., Honegger, F., Adkin, A. L., & Bloem, B. R. 2002, "Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man", *Journal of Physiology.*Vol.542, no. 2, pp. 643-663.

Andersen JL. Klitgaard H. Bangsbo J. Saltin B.1994 "Myosin heavy chain isoforms in single fibres from m. vastus lateralis of soccer players: effects of strength-training." *Acta Physiologica Scandinavica.* Vol. 150, no.1, pp.21-6.

Aniansson, A., Sperling, L., Rundgren, A., & Lehnberg, E. 1983, "Muscle function in 75-year-old men and women. A longitudinal study", *Scandinavian Journal of Rehabilitation Medicine – Supplementum,* vol.9, pp.92-102.

Aniansson, A., Zetterberg, C., Hedberg, M., & Henriksson, K. G. 1984, "Impaired muscle function with aging: A background factor in the incidence of fractures of the proximal end of the femur", *Clinical Orthopaedics & Related Research.*vol.191, pp. 193-201.

Aniansson, A., Hedberg, M., Henning, G. B., & Grimby, G. 1986, "Muscle morphology, enzymatic activity, and muscle strength in elderly men: a follow-up study", *Muscle and Nerve*, vol. 9, no. 7, pp. 585-591.

Aniansson, A., Grimby, G., & Hedberg, M. 1992, "Compensatory muscle fibre hypertrophy in elderly men", *Journal of Applied Physiology*, vol. 73, no. 3, pp. 812-816.

Astrand, P.O., Rodahl, K. (1986). Texbook of Work Physiology. McGraw Hill, New York.

Babault, N., Pousson, M., Michaut, A., & Van Hoecke, J. 2003, "Effect of quadriceps femoris muscle length on neural activation during isometric and concentric contractions", *Journal of Applied Physiology* .vol 94, no. 3, pp. 983-90.

Baker, M. R. & Baker, S. N. 2003, "The effect of diazepam on motor cortical oscillations and corticomuscular coherence studied in man", *Journal of Physiology,* vol. 546, no. 3, pp. 931-942.

Barry, B. K. & Carson, R. G. 2004, "The consequences of resistance training for movement control in older adults.", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 59, no.7, pp.730-54.

Bassey, E. J. & Short, A. H. 1990, "A new method for measuring power output in a single leg extension: feasibility, reliability and validity", *European Journal of Applied Physiology*, vol. 60, pp. 385-390.

Bassey, E. J., Fiatarone, M. A., O'Neill, E. F., Kelly, M., Evans, W. J., & Lipsitz, L. A. 1992, "Leg extensor power and functional performance in very old men and women", *Clinical Science*, vol. 82, pp. 321-327.

Bean, J. F., Kiely, D. K., Herman, S., Leveille, S. G., Mizer, K., Frontera, W. R., & Fielding, R. A. 2002, "The relationship between leg power and physical performance in mobility-limited older people", *Journal of the American Geriatrics Society, vol* 50, no. 3, pp. 461-467.

Begg, R. K. & Sparrow, W. A. 2000, "Gait characteristics of young and older individuals negotiating a raised surface: Implications for the prevention of falls", *Journal of Gerontology*, vol. 55A, no. 3, pp. M147-M154.

Behm, D., Power, K., & Drinkwater, E. 2001, "Comparison of interpolation and central activation ratios as measures of muscle inactivation", *Muscle & Nerve, vol.*24, no. 7, pp. 925-934.

Belanger, A.Y., McComas, A.J. 1981, "Extent of motor unit activation during effort." *Journal of Applied Physiology,* vol. 51, no. 5, pp.1131-1135

Bellew, J. W., Malone, T. R., Nitz, A. J., & Hart, A. L. 1998, "Gender specificity in the age-related decline of strength: concentric versus eccentric", *Isokinetics and Exercise Science*, vol. 7, no. 1, pp. 1-9.

Bellew, J. W. 2002, "The effect of strength training on control of force in older men and women", *Aging-Clinical & Experimental Research, vol.*14, no.1, pp. 35-41.

Bennie, S., Bruner, K., Dizon, A., Fritz, H., Goodman, B., & Peterson, S. 2003 "Measurements of balance: comparison of the Timed "Up and Go" test and Functional Reach test with the Berg Balance Scale", *Journal of Physical Therapy Science*, vol. 15, no. 2, pp. 93-97.

Bigland-Richie, B. Furbush, F., Woods, J.J., 1986. Neuromuscular transmission and muscle activation in human post-fatigue ischaemia. *Journal of Physiology*, vol. 377, no. 76P (Abstract).

Birmingham, A. T., Wharrad, H. J., & Williams, E. J. 1985, "The variation of finger tremor with age in man", *Journal of Neurology, Neurosurgery & Psychiatry*, vol. 48, no. 8, pp. 788-798.

Blake, A. J., Morgan, K., Bendall, M. J., Dallosso, H., Ebrahim, S. B. J., Arie, T. H. D., Fentem, P. H., & Bassey, E. J. 1988, "Falls by elderly people at home: Prevalence and associated factors", *Age & Ageing*, vol.17, no. 6, pp. 365-372.

Bonnefoy, M., Kostka, T., Arsac, L. M., Berthouze, S. E., & Lacour, J.-R. 1998, "Peak anaerobic power in elderly men", *European Journal of Applied Physiology & Occupational Physiology*.Vol.77, no. 1-2, pp. 182-188.

Bosco, C. & Komi, P. V. 1980, "Influence of aging on the mechanical behavior of leg extensor muscles", *European Journal of Applied Physiology & Occupational Physiology*, vol. 45, no. 2-3, pp. 209-219.

Brooks, S. V. & Faulkner, J. A. 1990, "Contraction-induced injury: Recovery of skeletal muscles in young and old mice", *American Journal of Physiology - Cell Physiology*.Vol.258, no. 3 27-3, p. C436-C442.

Brown, M., Sinacore, D. R., & Host, H. H. 1995, "The relationship of strength to function in the older adult", *The Journals of Gerontology, Series A*, vol. 50A, Special Issue, pp. 55-59.

Brown, W. F., Doherty, T. J., Chan, M., Andres, A., & Provost, S. M. 2000, "Human motor units in health and disease.", *Muscle & Nerve*, vol. Suppl 9, pp.S79-18.

Burnett, R. A., Laidlaw, D. H., & Enoka, R. M. 2000, "Co-activation of the antagonist muscle does notcovary with steadiness in old adults", *Journal of Applied Physiology*, vol. 89, pp. 61-71.

Campbell AJ., Buchner DM, Tilyard MW, Norton RN, Gardner MM, & Robertson MC 1997, "Randomised controlled trial of a general practice programme of home based exercise to prevent falls in elderly women", *British Medical Journal*, vol. 315, no. 7115, pp. 1065-1069.

Campbell, A. J., Spears, G. F., & Borrie, M. J. 1990, "Examination by logistic regression modelling of the variables which increase the relative risk of elderly women falling compared to elderly men", *Journal of Clinical Epidemiology*.Vol.43, no. 12, pp. 1415-1420.

Campbell, A. J., Robertson, M. C., Gardner, M. M., Norton, R. N., & Buchner, D. M. 1999, "Falls prevention over 2 years: a randomized controlled trial in women 80 years and older", *Age & Ageing*, vol .28, no. 6, pp. 513-518.

Capodaglio, P., Capodaglio, E. M., Ferri, A., Scaglioni, G., Marchi, A., & Saibene, F. 2005, "Muscle function and functional ability improves more in community-dwelling older women with a mixed-strength training programme", *Age & Ageing, vol.*34, no. 2, pp. 141-147.

Cham, R. & Redfern, M. S. 2001, "Lower extremity corrective reactions to slip events", *Journal of Biomechanics*, vol. 34, no. 11, pp. 1439-1445.

Chiu, A.Y.Y, Au-Yeung, S.S.Y., Lo, S.K., "A comparison of four functional tests in discriminating fallers from non-fallers in older people" *Disability and Rehabilitation*, vol 25, no.1, pp.45-50

Christakos, C. N. 1982, "A study of the muscle force waveform using a population stochastic model of skeletal muscle", *Biological Cybernetics*, vol.44, no. 2, pp. 91-106.

Christou, E. A. & Carlton, L. G. 2001, "Old adults exhibit greater motor output variability than young adults only during rapid discrete isometric contractions", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol 56, no. 12, pp B524-32.

Christou, E. A. & Carlton, L. G. 2002, "Age and contraction type influence motor output variability in rapid discrete tasks", *Journal of Applied Physiology*, vol. 93, no. 2, pp. 489-498.

Christou, E. A., Shinohara, M., & Enoka, R. M. 2003a, "Fluctuations in acceleration during voluntary contractions lead to greater impairment of movement accuracy in old adults", *Journal of Applied Physiology*, vol. 95, no. 1, pp. 373-384.

Christou, E. A., Yang, Y., & Rosengren, K. S. 2003b, "Taiji training improves knee extensor strength and force control in older adults", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 58, no. 8, pp. 763-766.

Christou, E.A., Jakobi, J.M., Critchlow, A., Fleshner, M., Enoka, R.M., 2004 "The 1-2 Hz oscillations in muscle force are exacerbated by stress, especially in older adults." *Journal of Applied Physiology*, vol. 97, pp. 225-235

Cole, K. J., Rotella, D. L., & Harper, J. G. 1998, "Tactile impairments cannot explain the effect of age on a grasp and lift task", *Experimental Brain Research*, vol.121, no. 3, pp. 263-269.

Connelly, D. M., Rice, C. L., Roos, M. R., & Vandervoort, A. A. 1999, "Motor unit firing rates and contractile properties in tibialis anterior of young and old men", *Journal of Applied Physiology*, vol. 87, no. 2, pp. 843-852.

Conway, B. A., Rosenberg J.R, Halliday, D. M., & Farmer, S. F. 1994. A frequency domain analysis of the relation between the electromyogram and physiological tremor. *Acta Physiologica Scandinavica,* vol. 151, no. 23A, P23.

Conway, B. A., Halliday, D. M., Farmer, S. F., Shahani, U., Maas, P., Weir, A. I., & Rosenberg, J. R. 1995, "Synchronization between motor cortex and spinal motoneuronal pool during the performance of a maintained motor task in man", *Journal of Physiology, vol.*489, no. 3, pp. 917-924.

Cress, M. E., Buchner, D. M., Questad, K. A., Esselman, P. C., DeLateur, B. J., & Schwartz, R. S. 1999, "Exercise: effects on physical functional performance in independent older adults", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences.*, vol. 54, no. 5, p. M242-M248.

Crosbie, W. J., Nimmo, M. A., Banks, M. A., Brownlee, M. G., & Meldrum, F. 1989, "Standing balance reponses in two populations of elderly women: a pilot study", *Archives of Physical Medicine and Rehabilitation,* vol. 70, no. 10, pp. 751-754.

Cunningham, D. A., Morrison, D., Rice, C. L., & Cooke, C. 1987, "Ageing and isokinetic plantar flexion", *European Journal of Applied Physiology & Occupational Physiology, vol.*56, no. 1, pp. 24-29.

D'Antona G, Pellegrino M A, Adami R, Rossi R, Carlizzi C N, Canepari M, Saltin B, Bottinelli R, 2003 "The effect of ageing and immobilization on structure and function of human skeletal muscle fibres" *Journal of Physiology,* vol. 552, 2, pp. 499-511

Danion, F. & Gallea, C. 2004, "The relation between force magnitude, force steadiness, and muscle co-contraction in the thumb during precision grip", *Neuroscience Letters, vol* 368, no. 2, pp176-80.

Danneskiold-Samsoe, B., Kofod, V., Munter, J., Grimby, G., Schnohr, P., & Jensen, G. 1984, "Muscle strength and functional capacity in 78-81 year old men and women", *European Journal of Applied Physiology* , vol. 52, pp. 310-314.

Darling, W. G., Cooke, J. D., & Brown, S. H. 1989, "Control of simple arm movements in elderly humans", *Neurobiology of Aging,* vol.10, no. 2, pp. 149-157.

Datta, A. K. & Stephens, J. A. 1990, "Synchronization of motor activity during voluntary contraction in man", *Journal of Physiology, vol.*422, pp. 397-419.

Datta, A. K., Farmer, S. F., & Stephens, J. A. 1991, "Central nervous pathways underlying synchronization of human motor unit firing studied during voluntary contractions", *J Physiol,* vol. 432, pp. 401-425.

Daubney, M. E. & Culham, E. G. 1999, "Lower extremity muscle force and balance performance in adults aged 65 years and older", *Physical Therapy,* vol. 79, no. 12, pp. 1177-1185.

Davis, J.W., Ross, P.D., Nevitt, M.C., Wasnich, R.D., "Risk factors for falls and for serious injuries on falling among older Japanese women in Hawaii" *Journal of the American Geriatrics Society*, vol 47, no. 7, pp.792-798.

de C. Hamilton, A., Jones, K.E, Wolpert, D.M., 2004, "The scaling of motor noise with muscle strength and motor unit number in humans". *Experimental Brain Research*, vol. 157, pp.417-430

De Luca, C.J., Le Fever, R.S., McCue, M.P., Xenakis, A.P., 1982, Control Scheme governing concurrently active human motor units during voluntary contractions. *Journal of Physiology*, vol. 329, pp. 129-142

De Luca, A., Mambrini M., Camerino D.C., 1990, "Changes in membrane ionic conductances and excitability characteristics of rat skeletal muscle during aging", *Pflugers Archives - European Journal of Physiology.*, vol. 415, pp. 642-4.

de Rekeneire, N., Visser, M., Peila, R., Nevitt, M. C., Cauley, J. A., Tylavsky, F. A., Simonsick, E. M., & Harris, T. B. 2003, "Is a fall just a fall: correlates of falling in healthy older persons. The Health, Aging and Body Composition Study", *Journal of the American Geriatrics Society.*, vol. 51, no. 6, pp. 841-846.

De Ruiter, C. J. & De Haan, A. 2001, "Similar effects of cooling and fatigue on eccentric and concentric force-velocity relationships in human muscle", *Journal of Applied Physiology* .vol.90, no. 6, pp. 2109-2116.

De Serres SJ, Enoka RM, & Kutzcher DV. 2002. Sensitivity of the force display influences performance on a force matching task. Joint meeting of the Canadian Physiological Society and Japanese Physiological Society, January 19-23, 2000. Lake Louise, Alberta, Canada. (Abstract).

DeVita, P. & Hortobagyi, T. 2000, "Age causes a redistribution of joint torques and powers during gait", *Journal of Applied Physiology*, vol. 88, pp. 1804-1811.

de Vreede, P. L., Samson, M. M., van Meeteren, N. L., Duursma, S. A., & Verhaar, H. J. 2005, "Functional-task exercise versus resistance strength exercise to improve daily function in older women: a randomized, controlled trial", *Journal - American Geriatrics Society*, vol. 53, no. 1, pp. 2-10.

Delbaere, K. & Bourgois 2003, "Age-related changes in concentric and eccentric muscle strength in the lower and upper extremity: a cross-sectional study", *Isokinetics & Exercise Science*, vol. 11, no. 3, pp. 145-151.

Delbono O 2002 "Molecular mechanisms and therapeutics of the deficit in specific force in ageing skeletal muscle". *Biogerontology* vol.3, pp.265-70

Deschenes, M. R. 2004, "Effects of Aging on Muscle Fibre Type and Size.", *Sports Medicine*, vol. 34, no. 12, pp. 809-824.

Duncan, P. W., Studenski, S., Chandler, J., & Prescott, B. 1992, "Functional reach: Predictive validity in a sample of elderly male veterans", *Journal of Gerontology*, vol. 47, no. 3, p. M93-M98.

Earles, D. R., Judge, J. O., & Gunnarsson, O. T. 2001, "Velocity training induces power-specific adaptations in highly functioning older adults", *Archives of Physical Medicine and Rehabilitation*, vol. 82, no. 7, pp. 872-878.

Edstrom, L. & Larsson, L. 1987, "Effects of age on contractile and enzyme-histochemical properties of fast- and slow-twitch single motor units in the rat", *Journal of Physiology*, vol. 392, pp. 129-145.

Ellaway, P. H. 1978, "Cumulative sum technique and its application to the analysis of peristimulus time histograms", *Electroencephalography & Clinical Neurophysiology, vol* 45, no. 2, pp. 302-4.

Eng, J. J., Winter, D. A., & Patla, A. E. 1994, "Strategies for recovery from a trip in early and late swing during human walking", *Experimental Brain Research, vol.*102, no. 2, pp. 339-349.

Enoka, R. M. 1996, "Eccentric contractions require unique activation strategies by the nervous system.", *Journal of Applied Physiology*, vol. 81, no. 6, pp. 2339-2346.

Enoka, R. M., Christou, E. A., Hunter, S. K., Kornatz, K. W., Semmler, J. G., Taylor, A. M., & Tracy, B. L. 2003, "Mechanisms that contribute to differences in motor performance between young and old adults", *Journal of Electromyography & Kinesiology*, vol. 13, no. 1, pp. 1-12.

Erim, Z., Beg, M. F., Burke, D. T., & DeLuca, C. J. 1999, "Effects of ageing on motor-unit control properties", *Journal of Neurophysiology*, vol. 82, no. 5, pp. 2081-2091.

Evans, W. J. 2000, "Exercise strategies should be designed to increase muscle power.", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences.*, vol. 55, no. 6, p. M309-M310.

Everitt B.S. 2003, *Medical statistics from A to Z.* Cambridge University Press.

Farmer, S. F., Bremner, F. D., Halliday, D. M., Rosenberg, J. R., & Stephens, J. A. 1993a, "The frequency content of common synaptic inputs to motoneurones studied during voluntary isometric contraction in man", *Journal of Physiology, vol.*470, pp. 127-155.

Farmer, S. F., Swash, M., Ingram, D. A., & Stephens, J. A. 1993b, "Changes in motor unit synchronization following central nervous lesions in man", *Journal of Physiology, vol.*463, pp. 83-105.

Farmer, S. F. 1998, "Rhythmicity, synchronization and binding in human and primate motor systems", *Journal of Physiology, vol.*509, no. 1, pp. 3-14.

Faulkner, J. A., Brooks, S. V., & Zerba, E. 1995, "Muscle atrophy and weakness with aging: Contraction-induced injury as an underlying mechanism", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences.*vol. 50, no. SPEC. ISSUE, pp. 124-129.

Faulkner, J. A. & Brooks, S. V. 1995, "Muscle fatigue in old animals. Unique aspects of fatigue in elderly humans.", *Advances in Experimental Medicine & Biology*, vol. 384, pp. 471-480.

Ferri, A., Scaglioni, G., Pousson, M., Capodaglio, P., Van Hoecke, J., & Narici, M. V. 2003, "Strength and power changes of the human plantar flexors and knee extensors in response to resistance training in old age", *Acta Physiologica Scandinavica*, vol 177, no.1, pp69-78.

Fiatarone, M. A., Marks, E. C., Ryan, N. D., Meredith, C. N., Lipsitz, L. A., & Evans, W. J. 1990, "High-intensity strength training in nonagenarians", *Journal of the American Medical Association*, vol. 263, no. 22, pp. 3029-3034.

Fiatarone, M. A., O'Neill E, & Ryan N. 1994. A randomised controlled trial of exercise and nutrition for physical frailty in the oldest old. New England Journal of Medicine vol 330, pp 1769-1775.

Fielding, R. A., LeBrasseur, N. K., Cuoco, A., Bean, J., Mizer, K., & Singh, M. A. F. 2002, "High-velocity resistance training increases skeletal muscle peak power in older women", *Journal of the American Geriatrics Society, vol* 50, no. 4, pp. 655-662.

Fisher, N. M., Pendergast, D. R., & Calkins, E. C. 1990, "Maximal isometric torque of knee extension as a function of muscle length in subjects of advancing age", *Archives of Physical Medicine & Rehabilitation, vol* 71, no. 10, pp729-34.

Fleming BE., Pendergast DR., & Wilson DR 1991, "A portable, easily performed muscle power test and its association with falls by elderly people", *Archives of Physical Medicine & Rehabilitation*, vol. 72, no. 11, pp. 886-889.

Foldvari, M., Clark, M., Laviolette, L. C., Bernstein, M. A., Kaliton, D., Castaneda, C., Pu, C. T., Hausdorff, J. M., Fielding, R. A., & Fiatarone, M. A. 2000, "Association of muscle power with functional status in community-dwelling elderly women", *Journal of Gerontology*, vol. 55A, no. 4, p. M192-M199.

Frontera, W. R., Meredith, C. N., O'Reilly, K. P., Knuttgen, H. G., & Evans, W. J. 1988, "Strength conditioning in older men: Skeletal muscle hypertrophy and improved function", *Journal of Applied Physiology, vol.*64, no. 3, pp. 1038-1044.

Frontera, W. R., Hughes V.A., Lutz K.J., & Evans W.J. 1991. A cross-sectional study of muscle strength and mass in 45- to 75-yr-old men and women. *Journal of Applied Physiology* vol. 71, no. 2, pp 644-650.

Frontera, W. R., Hughes, V. A., Fielding, R. A., Fiatarone, M. A., Evans, W. J., & Roubenoff, R. 2000a, "Ageing of skeletal muscle: a 12-yr longitudinal study", *Journal of Applied Physiology*, vol. 88, pp. 1321-1326.

Frontera, W. R., Suh, D., Krivickas, L. S., Hughes, V. A., Goldstein, R., & Roubenoff, R. 2000b, "Skeletal muscle fiber quality in older men and women", *American Journal of Physiology - Cell Physiology*, vol. 279, no. 3, p. C611-C618.

Galganski, M. E., Fuglevand, A. J., & Enoka, R. M. 1993, "Reduced control of motor output in a human hand muscle of elderly subjects during submaximal contractions", *Journal of Neurophysiology*, vol. 69, no. 6, pp. 2108-2115.

Gallagher, D., Visser, M., De Meersman, R. E., Sepulveda, D., Baumgartner, R. N., Pierson, R. N., Harris, T., & Heymsfield, S. B. 1997, "Appendicular skeletal muscle mass: Effects of age, gender, and ethnicity", *Journal of Applied Physiology*, vol. 83, no. 1, pp. 229-239.

Gardner, E. 1940. Decrease in Human neurones with age. *The Anatomical Record,* vol. 77, pp 529-536.

Gauchard, G. C., Tessier, A., Jeandel, C., & Perrin, P. 2003, "Improved muscle strength and power in elderly exercising regularly", *International Journal of Sports Medicine*, vol. 24, no. 1, pp. 71-74.

Gehlsen, G. M.,Whaley, M.H. 1990, "Falls in the elderly: part II, balance, strength, and flexibility", *Archives of Physical Medicine & Rehabilitation*, vol. 71, no. 10, pp. 739-741.

Giannini, F. Landoni, L. Meralla N. Minetti A. E. Narici M. V. 1989. Estimation of specific tension of human knee extensor muscles from in vitro physiological CSA and strength measurements. *Proceedings of the Physiological Society*, King's College London meeting, 15-16 December 1989, p 86P.

Goldspink, G. & Harridge, S. D. R. 2004, "Growth factors and muscle ageing", *Experimental Gerontology, vol.*39, no. 10, pp. 1433-1438.

Gonzalez, E., Messi, M. L., & Delbono, O. 2000, "The specific force of single intact extensor digitorum longus and soleus mouse muscle fibers declines with aging", *Journal of Membrane Biology*, vol. 178, no. 3, pp. 175-183.

Goodpaster, B. H., Carlson, C. L., Visser, M., Kelley, D. E., Scherzinger, A., Harris, T. B., Stamm, E., & Newman, A. B. 2001, "Attenuation of skeletal muscle and strength in the elderly: The Health ABC Study", *Journal of Applied Physiology, vol* 90, no. 6, pp 2157-65.

Graafmans, W. C., Ooms, M. E., Hofstee, H. M., Bezemer, P. D., Bouter, L. M., & Lips, P. 1996, "Falls in the elderly: a prospective study of risk factors and risk profiles", *American Journal of Epidemiology*, vol. 143, no. 11, pp. 1129-1136.

Grabiner MD, Koh TJ, Lundin TM, Jahnigen DW (1993) "Kinematics of recovery from a stumble." *Journal of Gerontology*, vol 48, no.3, pp. M97-102

Graves, A. E., Kornatz, K. W., & Enoka, R. M. 2000, "Older adults use a unique strategy to lift inertial loads with the elbow flexor muscles", *Journal of Neurophysiology*, vol. 83, no. 4, pp. 2030-2039.

Grindrod S., Round JM, & Rutherford OM 1987, "Type 2 fibre composition and force per cross-sectional area in the human quadriceps", *Journal of Physiology*, vol. 390, p. 154.

Grounds, M. D. 2002, "Reasons for the degeneration of ageing skeletal muscle: a central role for IGF-1 signalling", *Biogerontology*, vol. 3, pp. 19-24.

Hagbarth, K. E. & Young, R. R. 1979, "Participation of the stretch reflex in human physiological tremor", *Brain, vol.*102, no. 3, pp. 509-526.

Hakkinen, K. & Hakkinen, A. 1991, "Muscle cross-sectional area, force production and relaxation characteristics in women at different ages", *European Journal of Applied Physiology*, vol. 62, no. 6, pp. 410-414.

Hakkinen, K. & Hakkinen, A. 1995, "Neuromuscular adaptations during intensive strength training in middle-aged and elderly males and females", *Electromyography & Clinical Neurophysiology, vol.*35, no. 3, pp. 137-147.

Hakkinen, K., Alen, M., Kallinen, M., Izquierdo, M., Jokelainen, K., Lassila, H., Malkia, E., Kraemer, W. J., & Newton, R. U. 1998a, "Muscle CSA, force production, and activation of leg extensors during isometric and dynamic actions in middle-aged and elderly men and women", *Journal of Aging and Physical Activity, vol* 6, no. 3, pp. 232-247.

Hakkinen, K., Kallinen, M., Izquierdo, M., Jokelainen, K., Lassila, H., Malkia, E., Kraemer, W. J., Newton, R. U., & Alen, M. 1998b, "Changes in agonist-antagonist EMG, muscle CSA, and force during strength training in middle-aged and older people", *Journal of Applied Physiology, vol* 84, no.4, pp.1341-9.

Hakkinen K, Newton RU, Gordon SE, McCormick M 1998c "Changes in muscle morphology, electromographic activity, and force production characteristics during progressive strength training in young and older men" *Journals of Gerontology.Series A: Biological Sciences and Medical Sciences*, vol. 53A, no. 6, pp. 415-424

Hakkinen, K., Kraemer, W. J., Newton, R. U., & Alen, M. 2001, "Changes in electromyographic activity, muscle fibre and force production characteristics during heavy resistance/power strength training in middle-aged and older men and women", *Acta Physiologica Scandinavica.*, vol. 171, no. 1, pp. 51-62.

Hall, C. D., Woollacott, M. H., & Jensen, J. L. 1999. "Age-related changes in rate and magnitude of ankle torque development: implications for balance control", *Journals of Gerontology.Series A: Biological Sciences and Medical Sciences*, vol. 54A, no. 10, p. M507-M513.

Halliday, D. M., Conway, B. A., Farmer, S. F., & Rosenberg, J. R. 1999, "Load-independent contributions from motor-unit synchronization to human physiological tremor", *Journal of Neurophysiology*, vol. 82, no. 2, pp. 664-675.

Harridge, S. D. R., White, M. J., Carrington, C. A., Goodman, M., & Cummins, P. 1995, "Electrically evoked torque-velocity characteristics and isomyosin composition of the triceps surae in young and elderly men", *Acta Physiologica Scandinavica*, vol. 154, no. 4, pp. 469-477.

Harridge, S. D. R., Kryger, A., & Stensgaard, A. 1999, "Knee extensor strength, activation, and size in very elderly people following strength training", *Muscle and Nerve*, vol. 22, pp. 831-839.

Hasegawa O, Komiyama A, Kurita R, Ota S, Matsumoto M 1993 "Age related changes in nerve action potentials of median nerve – an analysis using intraneural neurography" *Clinical Neurology*, vol. 33, no. 10, pp. 1055-1058

Henderson, G., Tomlinson, B. E., & Gibson, P. H. 1980, "Cell counts in human cerebral cortex in normal adults throughout life using an image analysing computer", *Journal of the Neurological Sciences, vol.*46, no. 1, pp. 113-136.

Henneman, E. Mendell. L.M. 1981 "Functional organization of motoneuron pool and its inputs". In: *Handbook of Physiology, Motor Control, Part I*, edited by Brooks. V.B. Bethesda, MD: American Physiological Society, , pp. 423–507

Holliday, R. 2000, "Ageing research in the next century", *Biogerontology*, vol. 1, pp. 97-101.

Hortobagyi, T., Zheng, D., Weidner, M., Lambert, N. J., Westbrook, S., & Houmard, J. A. 1995, "The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 50, no. 6, p. B399-B406.

Hortobagyi, T., Tunnel, D., Moody, J., Beam, S., & DeVita, P. 2001, "Low or high intensity strength training partially restores impaired quadriceps force accuracy and steadiness in aged adults", *Journal of Gerontology*, vol. 56A, no. 1, pp. 38-47.

Howard, J. E., McGill, K. C., & Dorfman, L. J. 1988, "Age effects on properties of motor unit action potentials: ADEMG analysis", *Annals of Neurology*, vol. 24, no. 2, pp. 207-213.

Hruda, K. V., Hicks, A. L., & McCartney, N. 2003, "Training for muscle power in older adults: effects on functional abilities", *Canadian Journal of Applied Physiology.*, vol. 28, no. 2, pp. 178-189.

Hughes, S. M. & Schiaffino, S. 1999, "Control of muscle fibre size: a crucial factor in ageing", *Acta Physiologica Scandinavican*, vol. 167, pp. 307-312.

Hughes, V. A., Frontera, W. R., Wood, M., Evans, W. J., Dallal, G. E., Roubenoff, R., & Fiatarone Singh, M. A. 2001, "Longitudinal muscle strength changes in older adults: influence of muscle mass, physical activity, and health", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol* 56, no 5, pp B209-17.

Hunter, S. K., Thompson, M. W., & Adams, R. D. 2000, "Relationships among age-associated strength changes and physical activity level, limb dominance, and muscle group in women", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol* 55, no. 6, pp B264-73.

Hurley, B. F. 1995, "Age, gender, and muscular strength", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol* 50, no. SPEC. ISSUE, pp. 41-44.

Hurley, M. V., Rees, J., & Newham, D. J. 1998, "Quadriceps function, proprioceptive acuity and functional performance in healthy young, middle-aged and elderly subjects", *Age and Ageing*, vol. 27, pp. 55-62.

Ikezoe, T. & Asakawa 2003, "The relationship between quadriceps strength and balance to fall of elderly admitted to a nursing home", *Journal of Physical Therapy Science*, vol. 15, no. 2, pp. 75-79.

Izquierdo, M., Ibanez, J., Gorostiaga, E., Garrues, M., Zuniga, A., Anton, A., Larrion, J. L., & Hakkinen, K. 1999, "Maximal strength and power characteristics in isometric and dynamic actions of the upper and lower extremities in middle-aged and older men", *Acta Physiologica Scandinavica, vol.* 167, no. 1, pp 57-68.

Izquierdo, M., Hakkinen, K., Ibanez, J., Garrues, M., Anton, A., Zuniga, A., Larrion, J. L., & Gorostiaga, E. M. 2001, "Effects of strength training on muscle power and serum hormones in middle-aged and older men", *Journal of Applied Physiology*, vol. 90, no. 4, pp. 1497-1507.

Jakobi, J. M. & Rice, C. L. 2002, "Voluntary muscle activation varies with age and muscle group", *Journal of Applied Physiology*, vol. 93, pp. 457-462.

Jones, D.A. & Round J.M. 1990. Skeletal Muscle in Health and Disease, Manchester University Press, Manchester and New York.

Jonsson, E., Seiger, A., & Hirschfeld, H. 2005, "Postural steadiness and weight distribution during tandem stance in healthy young and elderly adults", *Clinical Biomechanics.*Vol. 20, no. 2, pp. 202-208.

Jonsson, E., Seiger, A., & Hirschfeld, H. 2004, "One-leg stance in healthy young and elderly adults: a measure of postural steadiness?", *Clinical Biomechanics.* vol. 19, no. 7, pp. 688-694.

Jozsi, A. C., Campbell, W. W., Joseph, L., Davey, S. L., & Evans, W. J. 1999, "Changes in power with resistance training in older and younger men and women", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol.*54, no. 11, p. M591-M596.

Jubrias, S. A., Odderson, I. R., Esselman, P. C., & Conley, K. E. 1997, "Decline in isokinetic force with age: muscle cross-sectional area and specific force", *Pflugers Archives - European Journal of Physiology*, vol. 434, pp. 246-253.

Kaelin-Lang A, Luft AR, Burstein AH, Sohn YH, Cohen LG 2002 "Modulation of human corticomotor excitability by somatosensory input" Journal of Physiology, vol 540, no. 2, pp. 623-633

Kakuda, N., Nagaoka, M., & Wessberg, J. 1999, "Common modulation of motor unit pairs during slow wrist movement in man", *Journal of Physiology, vol.* 520, no. 3, pp. 929-940.

Kamen, G., Sison, S. V., Du, C. C., & Patten, C. 1995, "Motor unit discharge behavior in older adults during maximal-effort contractions", *Journal of Applied Physiology*, vol. 79, no. 6, pp. 1908-1913.

Kamen, G. & Roy, A. 2000, "Motor-unit synchronization in young and elderly adults", *European Journal of Applied Physiology*, vol. 81, pp. 403-410.

Kapeller P, Schmidt R, Fazeka F 2004 "Quantitative MRI:evidence of usual ageing in the brain" *Topics in Magnetic Resonance Imaging*, vol 15, no. 6, pp. 343-347

Kawakami, Y., Abe, T., & Fukunaga, T. 1993, "Muscle-fiber pennation angles are greater in hypertrophied than in normal muscles", *Journal of Applied Physiology, vol.*74, no. 6, pp. 2740-2744.

Keen, D. A., Yue, G. H., & Enoka, R. M. 1994, "Training related enhancement in the control of motor output in elderly humans", *Journal of Applied Physiology*, vol. 77, no. 6, pp. 2648-2658.

Keenan, K.G., Farina, D.F., Maluf, K.S., Merletti, R., Enoka, R.M., 2005 "Influence of amplitude cancellation on the simulated surface electromyogram". *Journal of Applied Physiology*, vol. 98, pp.120-131

Kemoun, G., Thoumie, P., Boisson, D., & Guieu, J. D. 2002, "Ankle dorsiflexion delay can predict falls in the elderly", *Journal of Rehabilitation Medicine, vol.* 34, no. 6, pp. 278-283.

Kenny R.A., Rubenstein, L.Z. Martin, F.C., Tinetti, M.E. 2001 Guideline for the prevention of falls in older persons. *Journal of the American Geriatrics Society*, vol. 49, pp. 664-672.

Kent-Braun, J. A. & Le Blanc R. 1996. Quantitation of central activation failure during maximal voluntary contractions in humans. *Muscle Nerve*, vol. 19, pp 861-869.

Kent-Braun, J. A. & Ng, A. V. 1999, "Specific strength and voluntary muscle activation in young and elderly women and men", *Journal of Applied Physiology*, vol. 87, no. 1, pp. 22-29.

Kinoshita H. & Francis P.R. 1996, "A comparison of prehension force control in young and elderly individuals", *European Journal of Applied Physiology*, vol. 74, pp. 450-460.

Klein, C. S., Rice, C. L., & Marsh, G. D. 2001, "Normalized force, activation, and co-activation in the arm muscles of young and old men", *Journal of Applied Physiology*, vol. 91, pp. 1341-1349.

Klitgaard, H., Ausoni, S., & Damiani, E. 1989, "Sarcoplasmic reticulum of human skeletal muscle: age-related changes and effect of training", *Acta Physiologica Scandinavica*, vol. 137, no. 1, pp. 23-31.

Klitgaard, H., Mantoni, M., Schiaffino, S., Ausoni, S., Gorza, L., Laurent-Winter, C., Schnohr, P., & Saltin, B. 1990a, "Function, morphology and protein expression of ageing skeletal muscle: a cross-sectional study of elderly men with different training backgrounds", *Acta Physiol Scand*, vol. 140, no. 1, pp. 41-54.

Klitgaard, H., Zhou, M., Schiaffino, S., Betto, R., Salviati, G., & Saltin, B. 1990b, "Ageing alters the myosin heavy chain composition of single fibres from human skeletal muscle", *Acta Physiologica Scandinavica, vol.*140, no. 1, pp. 55-62.

Kornatz, K. W., Christou, E. A., & Enoka, R. M. 2002. Steadiness training reduces the variability of motor unit discharge rate in isometric and anisometric contractions performed by old adults. *Society for Neuroscience Abstracts,* vol 28, p. 665.10.

Kornatz, K. W., Enoka, R. M., Pascoe, M. A., Poston, B., Meyer, F. G., & Semmler, J. G. 2003. Correlated motor unit discharge is similar in young and old adults for slow concentric, but not eccentric, contractions of a hand muscle. *Society for Neuroscience Abstracts,* vol 29, p. 914.7.

Kornatz, K. W., Semmler J.G., Pascoe, M. A., Meyer, F. G., & Enoka, R. M. 2004. Correlated motor unit activity has only a minor influence on the fluctuations in acceleration during anisometric contractions. *Society for Neuroscience Abstracts* vol 30, p 188.12.

Kurugati U, Parker P, Rickards J, Tingley M, Sexsmith J 2005 "Bilateral isokinetic training reduces the bilateral leg strength deficit for both old and young adults" *European Journal of Applied Physiology* vol.94, pp. 175-179

Lafargue G. Paillard J. Lamarre Y. Sirigu A. 2003 "Production and perception of grip force without proprioception: Is there a sense of effort in deafferented subjects?". *European Journal of Neuroscience.* Vol. 17, no.12,pp 2741-2749.

Laforest, S., St Pierre, D. M., Cyr, J., & Gayton, D. 1990, "Effects of age and regular exercise on muscle strength and endurance", *European Journal of Applied Physiology & Occupational Physiology, vol* 60, no. 2, pp 104-11.

Laidlaw, D. H., Kornatz, K. W., Keen, D. A., Suzuki, S., & Enoka, R. M. 1999, "Strength training improves the steadiness of slow lengthening contractions performed by old adults", *Journal of Applied Physiology*, vol. 87, no. 5, pp. 1-10.

Laidlaw, D. H., Bilodeay, M., & Enoka, R. M. 2000, "Steadiness is reduced and motor-unit discharge is more variable in old adults", *Muscle and Nerve*, vol. 23, pp. 600-612.

Laidlaw, D. H., Hunter, S. K., & Enoka, R. M. 2002, "Nonuniform activation of the agonist muscle does not covary with index finger acceleration in old adults", *Journal of Applied Physiology*, vol. 93, no. 4, pp. 1400-1410.

Lamoureux E, Murphy, & Sparrow WA 2003, "The effects of improved strength on obstacle negotiation in community-living older adults", *Gait & Posture*, vol. 17, no. 3, pp. 273-283.

Lan, C., Lai, J. S., Chen, S. Y., & Wong, M. K. 1998, "12-month Tai Chi training in the elderly: its effect on health fitness", *Medicine and Science in Sports and Exercise*, vol. 30, no. 3, pp. 345-351.

Lanza, I. R., Towse, T. F., Caldwell, G. E., Wigmore, D. M., & Kent-Braun, J. A. 2003, "Effects of age on human muscle torque, velocity, and power in two muscle groups", *Journal of Applied Physiology, vol.*95, no. 6, pp. 2361-2369.

Lark, S. D., Buckley, J. G., Bennett, S., Jones, D., & Sargeant, A. J. 2003, "Joint torques and dynamic joint stiffness in elderly and young men during stepping down", *Clin Biomech (Bristol, Avon)*, vol. 18, no. 9, pp. 848-855.

Larsson, L., Grimby, G., & Karlsson, J. 1979, "Muscle strength and speed of movement in relation to age and muscle morphology", *Journal of Applied Physiology: Respiratory, Environmental & Exercise Physiology.*, vol. 46, no. 3, pp. 451-456.

Larsson, L. & Salviati, G. 1989, "Effects of age on calcium transport activity of sarcoplasmic reticulum in fast- and slow-twitch rat muscle fibres", *Journal of Physiology*, vol. 419, pp. 253-264.

Larsson, L., Li, X., & Frontera, W. R. 1997, "Effects of aging on shortening velocity and myosin isoform composition in single human skeletal muscle cells", *American Journal of Physiology*, vol. 272, no. 2:Pt 1, p. t-49.

Lazarus, J.-A. & Haynes, J. M. 1997, "Isometric pinch force control and learning in older adults", *Experimental Aging Research, vol.*23, no. 2, pp. 179-199.

Lee, L. W. & Kerrigan, D. C. 1999, "Identification of kinetic differences between fallers and nonfallers in the elderly", *American Journal of Physical Medicine and Rehabilitation*, vol. 78, no. 3, pp. 242-246.

Levy DI, Yea AI, Skelton DA, & Young A. 1994. Strength, power and functional ability. In: Passeri M (ed) *Geriatrics '94.International Association of gerontology European region clinical section. 85-93.* Rome, CIC Edizioni International.

Lexell, J., Taylor, C. C., & Sjostrom, M. 1988, "What is the cause of ageing atrophy?", *Journal of the Neurological Sciences* , vol. 84, pp. 275-294.

Lexell, J. & Downham, D. 1992, "What is the effect of ageing on type 2 muscle fibres", *Journal of the Neurological Sciences*, vol. 107, pp. 250-251.

Lexell, J., Downham, D. Y., Larsson, Y., Bruhn, E., & Morsing, B. 1995, "Heavy-resistance training in older Scandinavian men and women: short- and long-term effects on arm and leg muscles", *Scandinavian Journal of Medicine and Science in Sports*, vol. 5, no. 6, pp. 329-341.

Lindle, R. S., Metter, E. J., Lynch, N. A., Fleg, J. L., Fozard, J. L., Tobin, J., Roy, T. A., & Hurley, B. F. 1997, "Age and gender comparisons of muscle strength in 654 women and men aged 20-93 years", *Journal of Applied Physiology*, vol. 83, no. 5, pp. 1581-1587.

Lord SR., Williams, & Ward JA 1995, "The effect of a 12-month exercise trial on balance, strength, and falls in older women: a randomized controlled trial", *Journal of the American Geriatrics Society*, vol. 43, no. 11, pp. 1198-1206.

Lord, S. R., Clark, R. D., & Webster, I. W. 1991, "Physiological factors associated with falls in an elderly population", *Journal - American Geriatrics Society*, vol. 39, no. 12, pp. 1194-1200.

Lord, S. R., McLean, D., & Stathers, G. 1992, "Physiological factors associated with injurious falls in older people living in the community", *Gerontology, vol.*38, no. 6, pp. 338-346.

Lord, S. R., , H. A., Rogers, M. W., & Fitzpatrick, R. 1999, "Lateral stability, sensorimotor function and falls in older people", *Journal of the American Geriatrics Society*, vol. 47, pp. 1077-1081.

Lord, S. R. & Dayhew, J. 2001, "Visual risk factors for falls in older people", *Journal of the American Geriatrics Society*, vol. 49, no. 5, pp. 508-515.

Luukinen, H., Koski, K., Laippala, P., & Kivela, S.-L. 1995, "Predictors for recurrent falls among the home-dwelling elderly", *Scandinavian Journal of Primary Health Care, vol.*13, no. 4, pp. 294-299.

Ly, L. P. & Handelsman, D. J. 2002, "Muscle strength and ageing: Methodological aspects of isokinetic dynamometry and androgen administration", *Clinical & Experimental Pharmacology & Physiology, vol.*29, no. 1-2, pp. 39-47.

Lynch NA, Hurley BF, Fleg JL, Roy TA, Tobin JD, Fozard JL, Lindle RS, & Metter EJ 1999, "Muscle quality. I. Age-associated differences between arm and leg muscle groups", *Journal of Applied Physiology: Respiratory, Environmental & Exercise Physiology*, vol. 86, no. 1, pp. 188-194.

Macaluso A & De Vito G. 2003. Comparison between young and older women in explosive power output and its determinants during a single leg-press action after optimisation of load. *European journal of physiology* vol 90, pp 458-463.

Macaluso, A., Nimmo, M. A., Foster, J. E., Cockburn, M., McMillan, N. C., & De Vito, G. 2002, "Contractile muscle volume and agonist-antagonist co-activation account for differences in torque between young and older women", *Muscle & Nerve, vol* 25, no. 6, pp.858-63.

Macaluso, A. & De Vito, G. 2004, "Muscle strength, power and adaptations to resistance training in older people", *European Journal of Applied Physiology*, vol. 91, no. 4, pp. 450-472.

MacRae PG, Feltner ME, & Reinsch S. 1994. A 1 year exercise programme for older women: effects on fall, injuries, and physical performance. *Journal of Ageing and Physical Activity* vol 2, pp 127-142.

MacRae, P. G., Lacourse, M., & Moldavon, R. 1992, "Physical performance measures that predict faller status in community-dwelling older adults", *Journal of Orthopaedic and Sports Physical Therapy*, vol. 16, no. 3, pp. 123-128.

Maganaris C.N. 2001, "In vivo tendon mechanical properties in young adults and healthy elderly", obtained online in 2004:www.cheshire.mmu.ac.uk/exspsci/research/invivo.htm

Maki, B. E. & McIlroy, W. E. 1997, "The role of limb movements in maintaining upright stance: the "change-in-support" strategy", *Physical Therapy, vol* 77, no. 5, pp. 488-507.

Manini, T. M., Cook, S. B., Ordway, N. R., Ploutz-Snyder, R. J., & Ploutz-Snyder, L. L. 2005, "Knee extensor isometric unsteadiness does not predict functional limitation in older adults", *American Journal of Physical Medicine and Rehabilitation*, vol. 84, no. 2, pp. 112-121.

Margreth, A., Damiani, E., Bortoloso, E., & 1999, "Sarcoplasmic reticulum in aged skeletal muscle", *Acta Physiologica Scandinavican*, vol. 167, pp. 331-338.

Marsh, G. D., Paterson, D. H., Govindasamy, D., & Cunningham, D. A. 1999, "Anaerobic power of the arms and legs of young and older men", *Experimental Physiology*, vol. 84, no. 3, pp. 589-597.

Martin, J. C., Farrar, R. P., Wagner, B. M., & Spirduso, W. W. 2000, "Maximal power across the lifespan. [see comments]", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 55, no. 6, p. M311-M316.

McAuley, J. H., Rothwell, J. C., & Marsden, C. D. 1997, "Frequency peaks of tremor, muscle vibration and electromyographic activity at 10 Hz, 20 Hz and 40 Hz during human finger muscle contraction may reflect rhythmicities of central neural firing", *Experimental Brain Research*, vol. 114, no. 3, pp. 525-541.

McAuley, J. H. & Marsden, C. D. 2000, "Physiological and pathological tremors and rhythmic central motor control", *Brain*, vol. 123, no. 8, pp. 1545-1567.

McGibbon, C. A. & Krebs, D. E. 1999, "Effects of age and functional limitation on leg joint power and work during stance phase of gait", *Journal of Rehabilitation Research and Development*, vol. 36, no. 3, pp. 173-182.

McGibbon, C. A., Krebs, D. E., & Puniello, M. S. 2001, "Mechanical energy analysis identifies compensatory strategies in disabled elders' gait", *Journal of Biomechanics*, vol. 34, no. 4, pp. 481-490.

McKee, K. J., Orbell, S., & Radley, K. A. 1999, "Predicting perceived recovered activity in older people after a fall", *Disability and rehabilitation*, vol. 21, no. 12, pp. 555-562.

McNevin, N. Fahlman, M. M. Morgan A. L. Topp R. Boardley D. J. Horynak J. 2002 Predictors of functional ability in elderly men and women. *Medicine & Science in Sports & Exercise* vol 34, no. 5, Supplement 1, p S18.

Melzer, I., Benjuya, N., & Kaplanski, J. 2004. "Postural stability in the elderly: a comparison between fallers and non-fallers", *Age and Ageing*, vol. 33, no. 6, pp. 602-607.

Meuleman, J. R., Brechue, W. F., Kubilis, P. S., & Lowenthal, D. T. 2000, "Exercise training in the debilitated aged: strength and functional outcomes", *Archives of Physical Medicine & Rehabilitation.*, vol. 81, no. 3, pp. 312-318.

Miszko, T. A., Cress, M. E., Slade, J. M., Covey, C. J., Agrawal, S. K., & Doerr, C. E. 2003. "Effect of strength and power training on physical function in community-dwelling older adults", *Journals of Gerontology.Series A: Biological Sciences and Medical Sciences*, vol. 58A, no. 2, pp. 171-175.

Mochizuki, Y., Mizutani, T., & Takasu, T. 1995, "Amyotrophic lateral sclerosis with marked neurological asymmetry: clinicopathological study", *Acta Neuropathologica, vol* 90, no. 1, pp. 44-50.

Moreland, J. D., Richardson, J. A., Goldsmith, C. H., & Clase, C. M. 2004, "Muscle Weakness and Falls in Older Adults: A Systematic Review and Meta-Analysis", *Journal of the American Geriatrics Society*, vol. 52, no. 7, pp. 1121-1129.

Morganti, C. M., Nelson, M. E., Fiatarone, M. A., Dallal, G. E., Economos, C. D., Crawford, B. M., & Evans, W. J. 1995, "Strength improvements with 1 yr of progressive resistance training in older women", *Medicine and Science in Sports and Exercise*, vol. 27, no. 6, pp. 906-912.

Moritz, C.T., Barry, B.K., Pascoe, M.A., Enoka, R.M. 2005, "Discharge rate variability influences the variation in force fluctuations across the working range of a hand muscle", Journal of Neurophysiology, vol. 93, pp. 2449-2459.

Morris-Chatta, R., Buchner, D. M., de Lateur, B. J., Cress, M. E., & Wagner, E. H. 1994, "Isokinetic testing of ankle strength in older adults: assessment of inter-rater reliability and stability of strength over six months", *Archives of Physical Medicine and Rehabilitation*, vol. 75, no. 11, pp. 1213-1216.

Morse, C. I., Thom, J. M., Davis, M. G., Fox, K. R., Birch, K. M., & Narici, M. V. 2004, "Reduced plantarflexor specific torque in the elderly is associated with a lower activation capacity", *European Journal of Applied Physiology, vol.*92, no. 1-2, pp. 219-226.

Moxley Scarborough, D. Krebs, D.E., Harris, B.E., 1999 "Quadriceps strength and dynamic stability in elderly persons". Gait and Posture, vol 10, pp. 10-20

Mulrow CD, Kanten, & Gerety MB 1994, "A randomized trial of physical rehabilitation for very frail nursing home residents", *JAMA*, vol. 271, no. 7, pp. 519-524.

Murray KJ. Hill K. Phillips B. Waterston JA 2005. "A pilot study of falls risk and vestibular dysfunction in older fallers presenting to hospital Emergency Departments" *Disability and Rehabilitation* vol 27, no. 9pp. 499-506.

Narici MV, Hoppeler H, Kayser B, Landoni L Claassen H, Gavardi C, Conti M, Cerretelli P 1996 "Human quadriceps cross-sectional area, torque and neural activation during 6 months strength training" *Acta Physiologica Scandinavica*, Vol 157, No. 2, , pp. 175-186

Narici, M. 1999, "Human skeletal muscle architecture studied in vivo by non-invasive imaging techniques: functional significance and applications", *Journal of Electromyography and Kinesiology*, vol. 9, no. 2, pp. 97-103.

Narici, M. V., Maganaris, C. N., Reeves, N. D., & Capodaglio, P. 2003, "Effect of aging on human muscle architecture", *Journal of Applied Physiology*, vol. 95, no. 6, pp. 2229-2234.

Nelson, R. M., Soderberg, G. L., & Urbscheit, N. L. 1984, "Alteration of motor-unit discharge characteristics in aged humans", *Physical Therapy, vol.*64,no. 1, pp. 29-34.

Newton, R. U., Hakkinen, K., Hakkinen, A., McCormick, M., Volek, J., & Kraemer, W. J. 2002, "Mixed-methods resistance training increases power and strength of young and older men", *Medicine and Science in Sports and Exercise*, vol. 34, no. 8, pp. 1367-1375.

Nordstrom, M. A., Fuglevand, A. J., & Enoka, R. M. 1992, "Estimating the strength of common input to human motoneurons from the cross-correlogram", *Journal of Physiology, vol.*453, pp. 547-574.

O'Brien, K., Pickles, B., Culham, E., 1998, Clinical measures of balance in community-dwelling elderly female fallers and non-fallers. *Physiotherapy Canada*, vol. 50, no. 3, pp. 212-217

Okuzumi, H., Tanaka, A., Nakamura, T. 1996, "Age-related changes in the magnitude of postural sway in healthy women", Journal of Human Movement Studies, vol. 31, no. 5, pp. 249-261

O'Neill, D. E. T., Thayer, R. E., Taylor, A. W., Dzialoszynski, T. M., & Noble, E. G. 2000, "Effects of short-term resistance training on muscle strength and morphology in the elderly", *Journal of Aging & Physical Activity, vol.*8, no. 4, pp. 312-324.

Osternig LR, 1986 "Isokinetic dynamometry: implications for muscle testing and rehabilitation" *Exercise and Sport Sciences Reviews*, vol 14, pp 45-80.

Overend, T. J., Cunningham, D. A., Kramer, J. F., Lefcoe, M. S., & Paterson, D. H. 1992, "Knee extensor and knee flexor strength: cross-sectional area ratios in young and elderly men", *Journal of Gerontology*, vol 47, no. 6, pp. M204-10.

Owino, V., Yang, S. Y., & Goldspink, G. 2001, "Age-related loss of skeletal muscle function and the inability to express the autocrine form of insulin-like growth factor-1 (MGF) in response to mechanical overload", *FEBS Letters, vol* 505, no. 2, pp. 259-63.

Palacios, R. 2002, "The future of global ageing", *International Journal of Epidemiology*, vol. 31, no. 4, pp. 786-791.

Patten, C. & Kamen, G. 2000, "Adaptations in motor unit discharge activity with force control training in young and older human adults", *European Journal of Applied Physiology, vol.*83, no. 2-3, pp. 128-143.

Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. 2002, "Influence of lower extremity strength of healthy older adults on the outcome of an induced trip", *Journal of the American Geriatrics Societ.*, vol. 50, no. 2, pp. 256-262.

Pearson, S. J., Harridge, S. D., Grieve, D. W., Young, A., & Woledge, R. C. 2001, "A variable inertial system for measuring the contractile properties of human muscle", *Medicine & Science in Sports & Exercise, vol.* 33, no. 12, pp.2072-6.

Pearson, S. J., Cobbold, M., & Harridge, S. D. R. 2004, "Power output of the lower limb during variable inertial loading: A comparison between methods using single and repeated contractions", *European Journal of Applied Physiology, vol.*92, no. 1-2, pp. 176-181.

Person, R. S. & Kudina, L. P. 1972, "Discharge frequency and discharge pattern of human motor units during voluntary contraction of muscle", *Electroencephalography & Clinical Neurophysiology, vol* 32, no. 5, pp. 471-83.

Petrella, J. K., Kim, J.-S., Tuggle, S. C., Hall, S. R., & Bamman, M. M. 2005, "Age differences in knee extension power, contractile velocity, and fatigability", *Journal of Applied Physiology, vol.*98, no. 1, pp. 211-220.

Phillips, S. K., Bruce, S. A., & Woledge, R. C. 1991, "In mice, the muscle weakness due to age is absent during stretching", *Journal of Physiology*, vol. 437, pp. 63-70.

Phillips, S. K., Rook, K. M., Siddle, N. C., Bruce, S. A., & Woledge, R. C. 1993a, "Muscle weakness in women occurs at an earlier age than in men, but strength is preserved by hormone replacement therapy", *Clinical Science, vol* 84, no. 1, pp.95-8.

Phillips, S. K., Rowbury, J. L., Bruce, S. A., & Woledge, R. C. 1993b, "Muscle force generation and age: the role of sex hormones," in *Sensorimotor Impairment in the Elderly* ed. Stelmach, G.E. & Homberg, V., pp. 129-141. Kluwer Academic Publishers.

Phillips, S. K., Wiseman, R. W., Woledge, R. C., & Kushmerick, M. J. 1993c, "Neither changes in phosphorus metabolite levels nor myosin isoforms can explain the weakness in aged mouse muscle", *Journal of Physiology*, vol. 463, pp. 157-167.

Pijnappels, M., Bobbert, M. F., & van Dieen, J. H. 2005a, "How early reactions in the support limb contribute to balance recovery after tripping", *Journal of Biomechanics*, vol. 38, no. 3, pp. 627-634.

Pijnappels, M., Bobbert, M. F., & van Dieen, J. H. 2005b, "Control of support limb muscles in recovery after tripping in young and older subjects", *Experimental Brain Research*, vol. 160, no. 3, pp. 326-333.

Porter, M. M., Vandervoort, A. A., & Kramer, J. F. 1997, "Eccentric peak torque of the plantar and dorsiflexors is maintained in older women", *Journals of Gerontology.Series A, Biological Sciences & Medical Sciences*, vol. 52, no. 2, p. B125-B131.

Porter, M. M. M. 1995, "Concentric and eccentric knee extension strength in older and younger men and women", *Canadian Journal of Applied Physiology*, vol. 20, no. 4, pp. 429-439.

Poulin MJ., Cunningham DA, Kramer JF., Paterson DH, & Vandervoort AA 1992, "Eccentric and concentric torques of knee and elbow extension in young and older men", *Canadian Journal of Sport Sciences*, vol. 17, no. 1, pp. 3-7.

Pyka, G., Lindenberger, E., Charette, S., & Marcus, R. 1994, "Muscle strength and fiber adaptations to a year-long resistance training program in elderly men and women", *Journal of Gerontology, vol.*49, no. 1, p. M22-M27.

Quilliam, T. A. & Ridley, A. 1971, "The receptor community in the finger tip", *Journal of Physiology, vol* 216, no. 1, pp. 15P-17P.

Ranganathan, V. K., Siemionow, V., Sahgal, V., & Yue, G. H. 2001a, "Effects of aging on hand function", *Journal of the American Geriatrics Society*, vol. 49, no. 11, pp. 1478-1484.

Ranganathan, V. K., Siemionow, V., Sahgal, V., Liu, J. Z., & Yue, G. H. 2001b, "Skilled finger movement exercise improves hand function", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 56, no. 8, p. M518-M522.

Reeves, N. D., Maganaris, C. N., & Narici, M. V. 2003a, "Effect of strength training on human patella tendon mechanical properties of older individuals", *Journal of Physiology, vol.*548, no. 3, pp. 971-981.

Reeves, N. D., Narici, M. V., & Maganaris, C. N. 2003b, "Strength training alters the viscoelastic properties of tendons in elderly humans", *Muscle & Nerve, vol.*28, no. 1, pp. 74-81.

Reeves, N. D., Maganaris, C. N., & Narici, M. V. 2004a, "Ultrasonographic assessment of human skeletal muscle size", *European Journal of Applied Physiology*, vol. 91, no. 1, pp. 116-118.

Reeves, N. D., Narici, M. V., & Maganaris, C. N. 2004b, "Effect of resistance training on skeletal muscle-specific force in elderly humans", *Journal of Applied Physiology, vol.*96, no. 3, pp. 885-892.

Reeves, N. D., Narici, M. V., & Maganaris, C. N. 2004c, "In vivo human muscle structure and function: Adaptations to resistance training in old age", *Experimental Physiology, vol.*89, no. 6, pp. 675-689.

Reeves, N. D., Maganaris, C. N., & Narici, M. V. 2005, "Plasticity of dynamic muscle performance with strength training in elderly humans", *Muscle & Nerve, vol.*31, no. 3, pp. 355-364.

Reinsch, S., MacRae, P., Lachenbruch, P. A., & Tobis, J. S. 1992, "Attempts to prevent falls and injury: A prospective community study", *Gerontologist*, vol. 32, no. 4, pp. 450-456.

Renganathan, M., Messi, M. L., & Delbono, O. 1997, "Dihydropyridine receptor-ryanodine receptor uncoupling in aged skeletal muscle", *Journal of Membrane Biology*, vol. 157, no. 3, pp. 247-253.

Rhea, M. R., Alvar, B. A., Burkett, L. N., & Ball, S. D. 2003. "A meta-analysis to determine the dose response for strength development", *Medicine and Science in Sports and Exercise*, vol. 35, no. 3, pp. 456-464.

Ringsberg, K., Gerdham, P., Johansson, J., & Obrant, K. J. 1999, "Is there a relationship between balance, gait performance and muscular strength in 75 year old women", *Age and Ageing*, vol. 28, pp. 289-293.

Robertson, M. C., Devlin, N., Gardner, M. M., & Campbell, A. J. 2001, "Effectiveness and economic evaluation of a nurse delivered home exercise programme to prevent falls. 1: Randomised controlled trial", *BMJ.*, vol. 322, no. 7288, pp. 697-701.

Robinson, B. S., Gordon, J. M., Wallentine, S. W., & Visio, M. 2004, "Relationship between lower-extremity joint torque and the risk for falls in a group of community dwelling older adults", *Physiotherapy Theory & Practice, vol.*20, no. 3, pp. 155-173.

Roma, A. A., Chiarello, L. A., Barker, S. P., & Brenneman, S. K. 2001, "Examination and comparison of the relationships between strength, balance, fall history, and ambulatory function in older adults", *Issues on Aging, vol* 24, no. 2, pp. 21-30.

Roman, W. J., Fleckenstein, J., Gundersen, J. S., Alway, S. E., Peshock, R., & Gonyea, W. J. 1993, "Adaptations in the elbow flexors of elderly males after heavy-resistance training", *Journal of Applied Physiology, vol.*74, no. 2, pp. 750-754.

Roos, M. R., Rice, C. L., Connelly, D. M., & Vandervoort, A. A. 1999, "Quadriceps muscle strength, contractile properties, and motor unit firing rates in young and old men", *Muscle & Nerve, vol* 22, no. 8, pp. 1094-103.

Rose, D.J., Jones, C.J., Lucchese, N. 2002 "Predicting the probability of flalls in community-residing older adults using the 8 foot up and go: a new measure of functional mobility". *Journal of Ageing and Physical Activity*, vol. 10, no. 4, pp. 466-478

Rothwell J. Control of Human Voluntary Movement. 1994. London, Chapman and Hall.

Rothwell JC. Traub MM. Day BL. 1982 ."Manual motor performance in a deafferented man" .*Brain.* Vol. 105, no. 3, pp 515-542

Rubenstein, L. Z. & Josephson, K. R. 2002, "The epidemiology of falls and syncope", *Clinics in Geriatric Medicine, vol.* 18, no. 2, pp. 141-158.

Runge, M., Rittweger, J., Russo, C. R., Schiessl, H., & Felsenberg, D. 2004, "Is muscle power output a key factor in the age-related decline in physical performance? A comparison of muscle cross-section, chair-rising test and jumping power", *Clinical Physiology & Functional Imaging, vol.*24, no. 6, pp. 335-340.

Rutherford, O. M., Jones, D. A., & Newham, D. J. 1986, "Clinical and experimental application of the percutaneous twitch superimposition technique for the study of human muscle activation", *Journal of Neurology, Neurosurgery and Psychiatry*, vol. 49, pp. 1288-1291.

Sanes JN, Mauritz K-H, Everts EV, Dalakas MC, Chu A 1984 "Motor deficits in patients with large fibre sensory neuropathy" *Proceedings of the National academy of Sciences, USA*, vol.81, pp. 979-982

Scaf-Klomp, W., Sanderman, R., Ormel, J., & Kempen, G. I. J. M. 2003, "Depression in older people after fall-related injuries: A prospective study", *Age & Ageing, vol.*32, no. 1, pp. 88-94.

Scaglioni, G. & Ferri 2002, "Plantar flexor activation capacity and H reflex in older adults: adaptations to strength training", *Journal of Applied Physiology: Respiratory, Environmental & Exercise Physiology*, vol. 92, no. 6, pp. 2292-2302.

Schiffman, J. M. & Luchies, C. W. 2001, "The effects of motion on force control abilities", *Clinical Biomechanics*, vol. 16, no. 6, pp. 505-513.

Schiller, B. C., Casas, Y. G., Tracy, B. L., DeSouza, C. A., & Seals, D. R. 2000, "Age-related declines in knee extensor strength and physical performance in healthy Hispanic and Caucasian women", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol 55, no. 12, pp. B563-9.

Schillings, A. M., Van Wezel, B. M. H., Mulder, T., & Duysens, J. 2000, "Muscular responses and movement strategies during stumbling over obstacles", *Journal of Neurophysiology, vol.*83, no. 4, pp. 2093-2102.

Schmidt, R. F., Wahren, L. K., & Hagbarth, K.-E. 1990, "Multiunit neural responses to strong finger pulp vibration. I. Relationship to age", *Acta Physiologica Scandinavica, vol.*140, no. 1, pp. 1-10.

Schmied, A., Pagni, S., Sturm, H., & Vedel, J. P. 2000, "Selective enhancement of motoneurone short-term synchrony during an attention-demanding task", *Experimental Brain Research*, vol. 133, no. 3, pp. 377-390.

Schwendner, K. I., Mikesky, A. E., Holt, W. S., Jr., Peacock, M., & Burr, D. B. 1997, "Differences in muscle endurance and recovery between fallers and nonfallers, and between young and older women", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 52, no. 3, p. M155-M160.

Seidler-Dobrin, R. D., He, J., & Stelmach, G. E. 1998, "Co-activation to reduce variability in the elderly", *Motor Control, vol* 2, no. 4, pp.314-30.

Semmler, J. G. & Nordstrom, M. A. 1998, "Motor unit discharge and force tremor in skill- and strength-trained individuals", *Experimental Brain Research, vol.*119, no. 1, pp. 27-38.

Semmler, J. G., Steege, J. W., Kornatz, K. W., & Enoka, R. M. 2000a, "Motor-unit synchronization is not responsible for larger motor-unit forces in old adults", *Journal of Neurophysiology*, vol. 84, pp. 358-366.

Semmler J.G., Kutzscher, D. V., Enoka, R. M., & Zhou, S. 2000b Motor Unit Synchronisation is enhanced during slow shortening and lengthening contractions of the first dorsal interosseous muscle. *Society for Neuroscience Abstracts* vol 26, pp. 170.10.

Semmler, J. G., Enoka, R. M., Kern, D. S., & Kornatz, K. W. 2001. Motor unit synchronisation reduces the steadiness of anisometric contractions by a hand muscle. *Society for Neuroscience Abstracts,* vol 27, pp. 168.3.

Semmler, J. G., Kornatz, K. W., Dinenno, D. V., Zhou, S., & Enoka, R. M. 2002a, "Motor unit synchronisation is enhanced during slow lengthening contractions of a hand muscle", *Journal of Physiology, vol.*545, no. 2, pp. 681-695.

Semmler, J. G. 2002b, "Motor unit synchronization and neuromuscular performance. ", *Exercise & Sport Sciences Reviews, vol* 30, no. 1, pp. 8-14.

Semmler, J. G., Kornatz, K. W., & Enoka, R. M. 2003, "Motor-unit coherence during isometric contractions is greater in a hand muscle of older adults", *Journal of Neurophysiology*, vol. 90, no. 2, pp. 1346-1349.

Semmler, J. G., Sale, M. V., Meyer, F. G., & Nordstrom, M. A. 2004, "Motor-unit coherence and its relation with synchrony are influenced by training", *Journal of Neurophysiology, vol.*92, no. 6, pp. 3320-3331.

Seynnes, O., Hue, O., Ledrole, D., & Bernard, P. L. 2002, "Adapted physical activity in old age: effects of a low-intensity training program on isokinetic power and fatigability", *Aging-Clinical & Experimental Research.*, vol. 14, no. 6, pp. 491-498.

Seynnes, O., Singh, M. A. F., Hue, O., Pras, P., Legros, P., Bernard, PC., 2004, "Physiological and functional responses to low-moderate versus high intensity progressive resistance training in frail elders. ", *Journals Gerontology Medical Sciences*, vol. 59A 5, 503-509.

Shumway-Cook, A., Braur, S., Woollacott,M. 2000 "Predicting the probability for falls in community-dwelling older adults using the Timed Up and Go Test. *Physical Therapy*, vol. 80, no. 9, pp.896-903

Simoneau, G. G. & Krebs, D. E. 2000, "Whole-body momentum during gait: a preliminary study of non-fallers and frequent fallers", *Journal of Applied Biomechanics*, vol. 16, no. 1, pp. 1-13.

Sinaki, M., Nwaogwugwu, N. C., Phillips, B. E., & Mokri, M. P. 2001, "Effect of gender, age, and anthropometry on axial and appendicular muscle strength", *American Journal of Physical Medicine & Rehabilitation, vol* 80, no. 5, pp. 330-8.

Sipila, S., Multanen, J., Kallinen, M., Era,P. & Suominen, H. 1996, "Effects of strength and endurance training on isometric muscle strength and walking speed in elderly women", *Acta Physiologica Scandinavica*, vol. 156, pp. 457-464.

Sipila, S. & Suominen, H. 1996, "Quantitative ultrasonography of muscle: detection of adaptations to training in elderly women", *Archives of Physical Medicine and Rehabilitation*, vol. 77, no. 11, pp. 1173-1178.

Skalska, A., Ocetkiewicz, T., Zak, M., Grodzicki, T. 2004, "Influence of age on postural control parameters measured with a force platform" *New Medicine*, vol. 7, no. 1, pp.12-16.

Skelton, D. A., Greig, C. A., Davies, J. M., & Young, A. 1994, "Strength, power and related functional ability of healthy people aged 65-89 years", *Age and Ageing*, vol. 23, pp. 371-377.

Skelton, D. A., Young, A., Greig, C. A., & Malbut, K. E. 1995, "Effects of resistance training on strength, power and selected functional abilities of women aged 75 and older", *Journal of the American Geriatrics Society*, vol. 43, pp. 1081-1087.

Skelton, D. A., Phillips, S. K., Bruce, S. A., Naylor, C. H., & Woledge, R. C. 1999, "Hormone replacement therapy increases isometric muscle strength of adductor pollicis in post-menopausal women", *Clinical Science*, vol. 96, no. 4, pp. 357-364.

Skelton, D. A., Kennedy, J., & Rutherford, O. M. 2002, "Explosive power and asymmetry in leg muscle function in frequent fallers and non-fallers aged over 65", *Age & Ageing*, vol. 31, no. 2, pp. 119-125.

Sobolewski, J. L. Morgan, A. L. Fahlman M. M. McNevin N. Boardley D. Topp R. 2002. The relationship between tests of functional ability in the elderly. *Medicine and Science in Sports and Exercise,* vol. 34, no. 5, Supplement 1, pp. S18.

Sorkin, J. D., Muller, D. C., & Andres, R. 1999, "Longitudinal change in the heights of men and women: Consequential effects on body mass index", *Epidemiologic Reviews, vol.*21, no. 2, pp. 247-260.

Stackhouse, S. K., Stevens, J. E., Lee, S. C. K., Pearce, K. M., Snyder-Mackler, L., & Binder-Macleod, S. A. 2001. "Maximum voluntary activation in nonfatigued and fatigued muscle of young and elderly individuals", *Physical Therapy*, vol. 81, no. 5, pp. 1102-1109.

Startzell, J. K., Owens, D. A., Mulfinger, L. M., & Cavanagh, P. R. 2000, "Stair negotiation in older people: a review", *Journal - American Geriatrics Society*, vol. 48, no. 5, pp. 567-580.

Steffen, T. M., Hacker, T. A., & Mollinger, L. 2002, "Age- and Gender-related test performance in community-dwelling elderly people: Six minute walk test, Berg balance scale, timed up & go test, and gait speeds", *Physical Therapy*, vol. 82, no. 2, pp. 128-137.

Stevens, J. E., Binder-Macleod, S., & Snyder-Mackler, L. 2001, "Characterization of the human quadriceps muscle in active elders", *Archives of Physical Medicine and Rehabilitation*, vol. 82, no. 7, pp. 973-978.

Stevens, J. E., Stackhouse, S. K., Binder-Macleod, S. A., & Snyder-Mackler, L. 2003, "Are voluntary muscle activation deficits in older adults meaningful?", *Muscle & Nerve, vol* 27, no. 1, pp.:99-101.

Stiles, R. N. & Randall, J. E. 1967, "Mechanical factors in human tremor frequency", *Journal of Applied Physiology.*, vol. 23, no. 3, pp. 324-330.

Stokes, M., Hides, J., & Nassiri, D. K. 1997. "Musculoskeletal ultrasound imaging: diagnostic and treatment aid in rehabilitation", *Physical Therapy Reviews*, vol. 2, no. 2, pp. 73-92.

Straus, S. 2001, "Recent Advances - Geriatric Medicine", *British Medical Journal*, vol. 322, pp. 86-89.

Studenski S., Chandler, & Duncan PW 1991, "Postural responses and effector factors in persons with unexplained falls: results and methodologic issues", *Journal of the American Geriatrics Society*, vol. 39, no. 3, pp. 229-234.

Suzuki, T., Bean, J. F., & Fielding, R. A. 2001, "Muscle power of the ankle flexors predicts functional performance in community-dwelling older women", *Journal of the American Geriatrics Society, vol.*49, no. 9, pp. 1161-1167.

Swift, C. G. 2001, "Falls in late life and their consequences - Implementing effective services", *British Medical Journal*, vol. 322, no. 7290, pp. 855-857.

Takazawa, K., Arisawa, K., Honda, S., Shibata, Y., & Saito, H. 2003, "Lower-extremity muscle forces measured by a hand-held dynamometer and the risk of falls among day-care users in Japan: Using multinomial logistic regression analysis", *Disability & Rehabilitation*, vol. 25, no. 8, pp. 399-404.

Tang, P. F., Woollacott, M. H., & Chong, R. K. 1998, "Control of reactive balance adjustments in perturbed human walking: roles of proximal and distal postural muscle activity", *Experimental Brain Research, vol* 119, no. 2, pp.141-52.

Tang, P. F. & Woollacott, M. H. 1998, "Inefficient postural responses to unexpected slips during walking in older adults", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences*, vol. 53, no. 6, p. M471-M480.

Taniguchi Y 1997 "Lateral specificity in resistance training: the effect of bilateral and unilateral training" *European Journal of Applied Physiology and Occupational Physiology* vol: 75, pp. 144-50

Taylor, A. M., Christou, E. A., & Enoka, R. M. 2003, "Multiple features of motor-unit activity influence force fluctuations during isometric contractions", *Journal of Neurophysiology*, vol. 90, no. 2, pp. 1350-1361.

Thelen, D. G., Schultz, A. B., Alexander, N. B., & Ashton-Miller, J. A. 1996, "Effects of age on rapid ankle torque development", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol* 51, no. 5, pp. M226-32.

Thelen, D. G., Muriuki, M., James, J., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. 2000, "Muscle activities used by young and old adults when stepping to regain balance during a forward fall", *Journal of Electromyography & Kinesiology*, vol. 10, no. 2, pp. 93-101.

Tinetti, M. E., Doucette, J., Claus, E., & Marottoli, R. 1995. "Risk factors for serious injury during falls by older persons in the community", *Journal of the American Geriatrics Society*, vol. 43, no. 11, pp. 1214-1221.

Tinetti, M. E. S. 1988, "Risk factors for falls among elderly persons living in the community", *New England Journal of Medicine*, vol. 319, no. 26, pp. 1701-1707.

Tobis, J. S., Block, M., Steinhaus-Donham, C., Reinsch, S., Tamaru, K., & Weil, D. 1990, "Falling among the sensorially impaired elderly", *Archives of Physical Medicine & Rehabilitation, vol.* 71, no. 2, pp. 144-147.

Tomlinson, B. E. & Irving, D. 1977, "The numbers of limb motor neurons in the human lumbosacral cord throughout life", *Journal of the Neurological Sciences.*, vol. 34, no. 2, pp. 213-219.

Tracy, B. L., Ivey, F. M., Hurlbut, D., Martel, G. F., Lemmer, J. T., Siegel, E. L., Metter, E. J., Fozard, J. L., Fleg, J. L., & Hurley, B. F. 1999, "Muscle quality. II. Effects of strength training in 65- to 75-yr-old men and women", *Journal of Applied Physiology*, vol. 86, no. 1, pp. 195-201.

Tracy, B. L., Kern, D. S. Mehoudar P. D., Sehnert S. M., Byrnes W. C., Enoka R. M. 2001. Strength training does not improve the steadiness of muscle contractions in the knee extensors of older adults. *Medicine and Science in Spoprts and Exercise* vol. 33, no. 5, Supplement 1, pp. S254.

Tracy, B. L., Byrnes, W. C., & Enoka, R. M. 2004, "Strength training reduces force fluctuations during anisometric contractions of the quadriceps femoris muscles in old adults", *Journal of Applied Physiology, vol.*96, no. 4, pp. 1530-1540.

Tracy, B. L. Enoka R.M. 2002, "Older adults are less steady during submaximal isometric contractions with the knee extensor muscles", *Journal of Applied Physiology: Respiratory, Environmental & Exercise Physiology*, vol. 92, no. 3, pp. 1004-1012.

Trappe, T. A., Lindquist, D. M., & Carrithers, J. A. 2001, "Muscle-specific atrophy of the quadriceps femoris with aging", *Journal of Applied Physiology*, vol. 90, pp. 2070-2074.

Trappe, S., Gallagher, P., Harber, M., Carrithers, J., Fluckey, J., & Trappe, T. 2003, "Single muscle fibre contractile properties in young and old men and women", *Journal of Physiology, vol.*552, no. 1, pp. 47-58.

Trimble MH. Enoka RM. 1991. "Mechanisms underlying the training effects associated with neuromuscular electrical stimulation". *Physical Therapy*, vol. 71, no. 4), pp 273-80.

Vaillancourt, D. E. & Newell, K. M. 2003, "Aging and the time and frequency structure of force output variability", *Journal of Applied Physiology, vol.*94, no. 3, pp. 903-912.

Vaillancourt, D. E., Larsson, L., & Newell, K. M. 2003, "Effects of aging on force variability, single motor unit discharge patterns, and the structure of 10, 20, and 40 Hz EMG activity", *Neurobiology of Aging, vol.*24, no. 1, pp. 25-35.

Vallbo, A. B. & Wessberg, J. 1993, "Organization of motor output in slow finger movements in man", *Journal of Physiology, vol.*469, pp. 673-691.

Vandervoort, A. A. & McComas, A. J. 1986, "Contractile changes in opposing muscles of the human ankle joint with aging", *Journal of Applied Physiology*, vol. 61, no. 1, pp. 361-367.

Vandervoort, A. A., Kramer, J. F., & Wharram, E. R. 1990, "Eccentric knee strength of elderly females", *Journal of Gerontology*, vol. 45, no. 4, pp. 125-128.

Visser, M., Kritchevsky, S. B., Goodpaster, B. H., Newman, A. B., Nevitt, M., Stamm, E., & Harris, T. B. 2002, "Leg muscle mass and composition in relation to lower extremity performance in men and women aged 70 to 79: The health, aging and body composition study", *Journal of the American Geriatrics Society*, vol. 50, no. 5, pp. 897-904.

Walton, J. M., Roberts, N., & Whitehouse, G. H. 1997, "Measurement of the quadriceps femoris muscle using magnetic resonance and ultrasound imaging", *British Journal of Sports Medicine, vol.*31, no. 1, pp. 59-64.

Welford, A. T. 1984. Psychomotor Performance. *Annual Review of Gerontology and Geriatrics* vol. 4, pp. 237-273.

Welle, S., Totterman, S., & Thornton, C. 1996, "Effect of age on muscle hypertrophy induced by resistance training", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol* 51, no. 6, pp. M270-5.

Wessberg, J. & Vallbo, A. B. 1996, "Pulsatile motor output in human finger movements is not dependent on the stretch reflex", *Journal of Physiology, vol.*493, no. 3, pp. 895-908.

Wessberg, J. & Kakuda, N. 1999, "Single motor unit activity in relation to pulsatile motor output in human finger movements", *Journal of Physiology*, vol. 517, no. 1, pp. 273-285.

Whipple, R. H., Wolfson, L. I., & Amerman, P. M. 1987, "The relationship of knee and ankle weakness to falls in nursing home residents: An isokinetic study", *Journal of the American Geriatrics Society*, vol. 35, pp. 13-20.

Winegard, K. J., Hicks, A. L., Sale, D. G., & Vandervoort, A. A. 1996, "A 12-year follow-up study of ankle muscle function in older adults", *Journals of Gerontology.Series A, Biological Sciences & Medical Sciences*, vol. 51, no. 3, p. B202-B207.

Wojcik, L. A., Thelen, D. G., Schultz, A. B., Aston-Miller, J. A., & Alexander, N. B. 2000, "Age and gender difference in peak lower extremity joint torques and ranges of motion used during single-step balance recovery from a forward fall", *Journal of Biomechanics*, vol. 34, pp. 67-73.

Wolfson, L., Whipple, R., Derby, C., Judge, J., King, M., Amerman, P., Schmidt, J., & Smyers, D. 1996, "Balance and strength training in older adults: intervention gains and Tai Chi maintenance", *Journal - American Geriatrics Society*, vol. 44, no. 5, pp. 498-506.

Wolfson, L. I., Whipple, R., & Amerman, P. 1985, "Gait and balance in the elderly. Two functional capacities that link sensory and motor ability to falls", *Clinics in Geriatric Medicine, vol.*1, no. 3, pp. 649-660.

Yan, J. H. 1999, "Tai chi practice reduces movement force variability for seniors", *Journals of Gerontology Series A-Biological Sciences & Medical Sciences, vol* 54, no. 12, pp. M629-34.

Yao, W., Fuglevand, A. J., & Enoka, R. M. 2000, "Motor-unit synchronization increases EMG amplitude and decreases force steadiness of simulated contractions", *Journal of Neurophysiology*, vol. 83, no. 1, pp. 441-452.

Young, A., Stokes, M., & Crowe, M. 1984, "Size and strength of the quadriceps muscles of old and young women", *European Journal of Clinical Investigation.*, vol. 14, no. 4, pp. 282-287.

Young, A., Stokes, M., & Crowe, M. 1985, "The size and strength of the quadriceps muscles of old and young men", *Clinical Physiology.*, vol. 5, no. 2, pp. 145-154

Appendix 1: Reliability of measures for the Kin Com isokinetic dynamometer

The following unpublished CoV values were collected by two colleagues at KCL for the purposes of another study.

6 young subjects were familiarised with the test procedures, and practice efforts were performed. Subjects then performed isometric quadriceps contractions at 90, 70 and 30°, isometric hamstring contractions at 20, 30 and 70°, and concentric and eccentric contractions at $60°.sec^{-1}$ and $150°.sec^{-1}$ on the quadriceps and hamstrings using their less dominant leg. During the session this was repeated twice, and the best value was recorded. The session was repeated 6 times, with between 2 and 7 days between each test, at the same time of day. Each subject's co-efficient of variation was derived from the six repeated sessions. The mean CoV of all 6 subjects were as shown in table A1.1

Test	n	CoV (%)
Isom Quads 90°	6	4.1%
Isom Quads 70°	6	9.7%
Isom Quads 30°	6	13.6%
Isom Hams 70°	6	12.3%
Isom Hams 30°	6	8.4%
Isom Hams 20°	6	8.4%
Quads concentric $60°s^{-1}$	6	4.6%
Hams concentric $60°s^{-1}$	6	16.7%
Quads concentric $150°s^{-1}$	6	5.9%
Hams concentric $150°s^{-1}$	6	3.6%
Quads eccentric $60°s^{-1}$	6	6.5%
Hams eccentric $60°s^{-1}$	6	2.2%
Quads eccentric $150°s^{-1}$	6	14.3%
Hams eccentric $150°s^{-1}$	6	7.7%

TableA1.1 CoV of repeated measures for isometric and isokinetic measurements.

Only young subjects were tested because of difficulties recruiting older subjects. However, it has been shown that older subjects have improved reproducibility of isokinetic measures (Ly and Handelsman 2002) so the data for the younger subjects may mark the lower limit of reproducibility expected in all subjects. This will allow conservative estimation of the effects of reproducibility on results.

Appendix 2: Strength data normalised to body mass and square of corrected height

Young v older subjects normalised for weight (N/kg)

	Young		Non faller	
	mean	SE	mean	SE
Quadriceps 90s	7.82	0.28	5.18	0.28
Quadriceps 90we	6.99	0.26	4.53	0.26
Quadriceps 80s	8.08	0.28	5.58	0.30
Quadriceps 80we	7.32	0.25	4.95	0.26
Quadriceps 70s	7.94	0.25	5.62	0.26
Quadriceps 70we	7.27	0.24	5.06	0.25
Quadriceps 60s	7.29	0.22	5.28	0.23
Quadriceps 60we	6.53	0.22	4.78	0.23
Quadriceps 50s	6.41	0.20	4.81	0.21
Quadriceps 50we	5.68	0.19	4.29	0.20
Quadriceps 40s	5.36	0.17	4.14	0.17
Quadriceps 40we	4.81	0.16	3.64	0.17
Quadriceps 30s	4.50	0.14	3.44	0.14
Quadriceps 30we	4.02	0.13	2.98	0.13
Quadriceps 20s	3.24	0.12	2.45	0.12
Quadriceps 20we	3.25	0.11	2.39	0.12
Hamstrings 90s	2.83	0.11	1.74	0.10
Hamstrings 90we	2.37	0.10	1.47	0.10
Hamstrings 80s	3.05	0.09	1.90	0.10
Hamstrings 80we	2.62	0.09	1.60	0.09
Hamstrings 70s	3.16	0.10	2.02	0.10
Hamstrings 70we	2.76	0.08	1.70	0.09
Hamstrings 60s	3.31	0.10	2.11	0.11
Hamstrings 60we	2.87	0.09	1.80	0.09
Hamstrings 50s	3.41	0.10	2.19	0.11
Hamstrings 50we	2.96	0.10	1.82	0.10
Hamstrings 40s	3.47	0.11	2.24	0.11
Hamstrings 40we	3.04	0.11	1.89	0.11
Hamstrings 30s	3.60	0.11	2.36	0.12
Hamstrings 30we	3.03	0.12	1.98	0.12
Hamstrings 20s	3.54	0.12	2.37	0.13
Hamstrings 20we	2.97	0.13	1.98	0.13
S quadriceps C 50	7.05	0.25	4.45	0.30
S quadriceps C 150	5.25	0.17	3.48	0.20
S quadriceps E 50	9.48	0.33	6.92	0.38
S quadriceps E 150	8.84	0.29	6.80	0.33
We quadriceps C 50	6.20	0.25	3.68	0.29
We quadriceps C 150	4.73	0.18	2.79	0.20
We quadriceps E 50	8.06	0.29	5.90	0.34
We quadriceps E 150	7.62	0.27	5.83	0.31
S hamstrings C 50	3.46	0.12	1.88	0.14
S hamstrings C 150	3.21	0.10	2.05	0.11
S hamstrings E 50	4.06	0.14	2.96	0.16
S hamstrings E 150	3.85	0.11	3.03	0.12
We hamstrings C 50	2.84	0.11	1.41	0.13
We hamstrings C 150	2.76	0.10	1.61	0.11
We hamstrings E 50	3.32	0.12	2.43	0.14
We hamstrings E 150	3.30	0.10	2.59	0.11

Table A2.1 Data of young and older subjects for strength normalised to body mass. S=strong, We=weak, CON=concentric, ECC=eccentric. For isometric, numbers refer to angle, for isokinetic numbers refer to speed.

Young v older subjects normalised for square of corrected height (N/m^2)

	Young		Non faller	
	mean	SE	mean	SE
Quadriceps 90s	180.60	7.09	121.59	6.41
Quadriceps 90w	162.20	6.52	106.82	5.89
Quadriceps 80s	187.85	7.18	131.15	6.98
Quadriceps 80w	167.95	6.38	116.04	6.20
Quadriceps 70s	183.38	6.43	131.67	6.25
Quadriceps 70w	166.52	6.17	118.42	6.00
Quadriceps 60s	167.24	5.58	123.52	5.42
Quadriceps 60w	147.45	5.40	111.79	5.24
Quadriceps 50s	146.78	5.05	112.54	4.90
Quadriceps 50w	128.72	4.78	99.94	4.64
Quadriceps 40s	122.91	4.42	96.47	4.25
Quadriceps 40w	109.28	4.16	84.54	4.01
Quadriceps 30s	103.44	3.60	80.07	3.40
Quadriceps 30w	91.17	3.25	69.09	3.07
Quadriceps 20s	75.08	3.01	56.68	2.89
Quadriceps 20w	76.25	2.92	55.20	2.80
Hamstrings 90s	66.13	2.23	40.83	2.01
Hamstrings 90w	56.07	2.10	34.54	1.89
Hamstrings 80s	70.45	1.99	44.65	1.95
Hamstrings 80w	60.11	1.94	37.56	1.89
Hamstrings 70s	73.38	2.19	47.60	2.15
Hamstrings 70w	64.83	2.00	40.00	1.96
Hamstrings 60s	76.99	2.44	49.84	2.39
Hamstrings 60w	67.04	2.24	42.33	2.19
Hamstrings 50s	79.21	2.60	51.68	2.55
Hamstrings 50w	68.68	2.61	42.96	2.56
Hamstrings 40s	80.74	2.87	53.00	2.77
Hamstrings 40w	71.12	2.90	44.68	2.80
Hamstrings 30s	83.32	3.10	56.08	3.04
Hamstrings 30w	70.67	3.15	47.13	3.09
Hamstrings 20s	83.71	3.38	56.34	3.28
Hamstrings 20w	70.23	3.43	47.11	3.33
S quadriceps C 50	162.67	6.69	103.09	7.23
S quadriceps C 150	120.44	4.74	80.84	4.97
S quadriceps E 50	216.73	8.30	159.44	8.98
S quadriceps E 150	200.43	7.33	157.06	7.69
We quadriceps C 50	140.87	6.46	84.95	6.98
We quadriceps C 150	107.63	4.62	64.79	4.84
We quadriceps E 50	183.55	7.30	136.52	7.90
We quadriceps E 150	173.33	7.03	135.40	7.37
S hamstrings C 50	77.86	3.06	43.93	3.46
S hamstrings C 150	74.11	2.49	47.23	2.65
S hamstrings E 50	92.89	3.62	68.19	3.87
S hamstrings E 150	87.88	2.77	69.99	2.87
We hamstrings C 50	65.48	2.96	33.31	3.24
We hamstrings C 150	63.48	2.56	37.28	2.72
We hamstrings E 50	76.39	3.29	56.61	3.51
We hamstrings E 150	75.24	2.54	60.07	2.63

Table A2.2 Data of young and older subjects for strength normalised to square of corrected height. S=strong, We=weak, CON=concentric, ECC=eccentric. For isometric, numbers refer to angle, for isokinetic angles refer to speed.

Faller v non-faller subjects normalised for body mass (N/kg)

	Fallers		Non faller	
	mean	SE	mean	SE
Quadriceps 90s	3.78	0.47	4.34	0.29
Quadriceps 90w	3.88	0.33	4.53	0.26
Quadriceps 80s	4.65	0.38	5.58	0.30
Quadriceps 80w	4.07	0.34	4.95	0.26
Quadriceps 70s	4.60	0.36	5.62	0.26
Quadriceps 70w	4.10	0.34	5.06	0.25
Quadriceps 60s	4.41	0.28	5.28	0.23
Quadriceps 60w	3.91	0.29	4.78	0.23
Quadriceps 50s	3.89	0.26	4.81	0.21
Quadriceps 50w	3.40	0.25	4.29	0.20
Quadriceps 40s	3.12	0.22	4.14	0.17
Quadriceps 40w	2.71	0.22	3.64	0.17
Quadriceps 30s	2.73	0.18	3.44	0.14
Quadriceps 30w	2.39	0.17	2.98	0.13
Quadriceps 20s	2.01	0.16	2.45	0.12
Quadriceps 20w	1.89	0.15	2.39	0.12
Hamstrings 90s	1.31	0.14	1.74	0.10
Hamstrings 90w	1.07	0.13	1.47	0.10
Hamstrings 80s	1.41	0.12	1.90	0.10
Hamstrings 80w	1.16	0.12	1.60	0.09
Hamstrings 70s	1.44	0.14	2.02	0.10
Hamstrings 70w	1.16	0.12	1.70	0.09
Hamstrings 60s	1.63	0.14	2.11	0.11
Hamstrings 60w	1.30	0.12	1.80	0.09
Hamstrings 50s	1.63	0.14	2.19	0.11
Hamstrings 50w	1.34	0.13	1.82	0.10
Hamstrings 40s	1.58	0.15	2.24	0.11
Hamstrings 40w	1.30	0.15	1.89	0.11
Hamstrings 30s	1.74	0.15	2.36	0.12
Hamstrings 30w	1.33	0.16	1.98	0.12
Hamstrings 20s	1.65	0.18	2.37	0.13
Hamstrings 20w	1.25	0.18	1.98	0.13
Dorsi 30s	2.71	0.13	3.25	0.10
Dorsi 30w	2.12	0.15	2.76	0.11
Dorsi 20s	2.37	0.14	2.82	0.10
Dorsi 20w	1.79	0.14	2.34	0.10
Dorsi 10s	1.98	0.16	2.07	0.10
Dorsi 10w	1.47	0.17	1.50	0.11
Dorsi 0s	1.20	0.19	1.14	0.17
Dorsi 0w	0.72	0.18	0.65	0.16
Plantar 30s	2.20	0.34	3.16	0.23
Plantar 30w	1.59	0.32	2.44	0.21
Plantar 20s	3.64	0.43	5.21	0.31
Plantar 20w	2.73	0.41	4.24	0.30
Plantar 10s	5.35	0.53	7.34	0.39
Plantar 10w	4.30	0.53	6.03	0.38
Plantar 0s	6.72	0.74	9.04	0.53
Plantar 0w	5.79	0.73	7.55	0.53

Table A2.3 Data of faller and non-faller subjects for isometric strength normalised to body mass. Dorsi=dorsiflexors, Plantar=plantarflexors, S=strong, We=weak, C=concentric, E=eccentric. Numbers refer to angle

	Fallers		Non faller	
	mean	SE	mean	SE
S dorsi C 50	1.24	0.11	1.31	0.08
S dorsi C 150	1.06	0.08	0.97	0.06
S dorsi E 50	3.14	0.24	3.84	0.17
S dorsi E 150	2.99	0.25	3.13	0.16
We dorsi C 50	1.01	0.10	1.07	0.07
We dorsi C 150	0.88	0.07	0.82	0.05
We dorsi E 50	2.87	0.25	3.33	0.17
We dorsi E 150	2.74	0.25	2.70	0.16
S plantar C 50	2.86	0.40	3.54	0.30
S plantar C 150	2.16	0.25	2.47	0.19
S plantar E 50	7.06	0.66	8.76	0.49
S plantar E 150	6.18	0.60	7.02	0.45
We plantar C 50	2.19	0.37	2.96	0.27
We plantar C 150	1.59	0.21	1.98	0.16
We plantar E 50	5.58	0.69	7.33	0.51
We plantar E 150	4.61	0.60	5.71	0.45
S quadriceps C 50	3.58	0.35	4.45	0.30
S quadriceps C 150	3.09	0.24	3.48	0.20
S quadriceps E 50	6.10	0.46	6.92	0.38
S quadriceps E 150	5.93	0.40	6.80	0.33
We quadriceps C 50	2.84	0.34	3.68	0.29
We quadriceps C 150	2.61	0.25	2.79	0.20
We quadriceps E 50	5.03	0.41	5.90	0.34
We quadriceps E 150	5.09	0.38	5.83	0.31
S hamstrings C 50	1.44	0.16	1.88	.14
S hamstrings C 150	1.67	0.13	2.05	0.11
S hamstrings E 50	2.39	0.19	2.96	0.16
S hamstrings E 150	2.69	0.15	3.03	0.12
We hamstrings C 50	1.39	0.14	1.39	0.14
We hamstrings C 150	1.23	0.14	1.61	0.11
We hamstrings E 50	1.76	0.17	2.43	0.14
We hamstrings E 150	2.26	0.14	2.59	0.11

Table A2.4 Data of faller and non-faller subjects for isokinetic strength normalised to body mass. Dorsi=dorsiflexors, Plantar=plantarflexors, S=strong, We=weak, C=concentric, E=eccentric. Numbers refer to speed.

Faller v non-faller subjects normalised for square of corrected height (N/m^2)

	Fallers		Non faller	
	mean	SE	mean	SE
Quadriceps 90s	111.28	8.22	121.59	6.41
Quadriceps 90we	96.76	7.56	106.82	5.89
Quadriceps 80s	115.00	8.91	131.15	6.98
Quadriceps 80we	100.24	7.92	116.04	6.20
Quadriceps 70s	114.01	8.51	131.67	6.25
Quadriceps 70we	101.22	8.16	118.42	6.00
Quadriceps 60s	108.59	6.83	123.52	5.42
Quadriceps 60we	95.90	6.61	111.79	5.24
Quadriceps 50s	96.34	6.18	112.54	4.90
Quadriceps 50we	83.66	5.85	99.94	4.64
Quadriceps 40s	77.99	5.56	96.47	4.25
Quadriceps 40we	68.00	5.23	84.54	4.01
Quadriceps 30s	67.77	4.38	80.07	3.40
Quadriceps 30we	59.40	3.95	69.09	3.07
Quadriceps 20s	49.61	3.83	56.68	2.89
Quadriceps 20we	47.16	3.71	55.20	2.80
Hamstrings 90s	32.77	2.64	40.83	2.01
Hamstrings 90we	27.07	2.49	34.54	1.89
Hamstrings 80s	35.45	2.52	44.65	1.95
Hamstrings 80we	29.19	2.45	37.56	1.89
Hamstrings 70s	36.40	2.94	47.60	2.15
Hamstrings 70we	29.33	2.69	40.00	1.96
Hamstrings 60s	40.72	3.12	49.84	2.39
Hamstrings 60we	32.40	2.86	42.33	2.19
Hamstrings 50s	40.75	3.28	51.68	2.55
Hamstrings 50we	33.25	3.29	42.96	2.56
Hamstrings 40s	39.53	3.80	53.00	2.77
Hamstrings 40we	32.68	3.84	44.68	2.80
Hamstrings 30s	43.50	3.91	56.08	3.04
Hamstrings 30we	33.25	3.97	47.13	3.09
Hamstrings 20s	41.52	4.57	56.34	3.28
Hamstrings 20we	31.93	4.64	47.11	3.33
Dorsi 30s	67.94	3.46	76.47	2.54
Dorsi 30we	52.28	3.56	64.41	2.60
Dorsi 20s	59.02	3.42	66.43	2.46
Dorsi 20we	44.68	3.37	55.38	2.43
Dorsi 10s	47.87	3.64	48.57	2.42
Dorsi 10we	36.65	3.88	35.36	2.58
Dorsi 0s	*29.30*	*4.87*	26.17	4.44
Dorsi 0we	18.73	4.25	15.25	3.88
Plantar 30s	56.61	8.21	74.09	5.45
Plantar 30we	41.23	7.49	56.73	4.98
Plantar 20s	92.33	10.10	122.32	7.30
Plantar 20we	69.53	9.36	98.87	6.76
Plantar 10s	135.38	12.29	172.91	8.88
Plantar 10we	109.14	11.65	141.09	8.41
Plantar 0s	169.44	16.62	213.56	11.91
Plantar 0we	145.80	16.19	177.31	11.60

Table A2.5 Data of faller and non-faller subjects for isometric strength normalised to square of corrected heightt. Dorsi=dorsiflexors, Plantar=plantarflexors S=strong, We=weak. Numbers refer to angle.

	Fallers		Non faller	
	mean	SE	mean	SE
S dorsi C 50	30.28	2.48	30.54	1.78
S dorsi C 150	25.93	1.68	22.50	1.16
S dorsi E 50	75.53	5.33	87.53	3.75
S dorsi E 150	71.88	5.49	72.66	3.55
We dorsi C 50	24.40	2.39	24.72	1.71
We dorsi C 150	21.71	1.29	19.11	0.89
We dorsi E 50	69.20	5.36	75.83	3.77
We dorsi E 150	66.64	5.53	62.42	3.57
S plantar C 50	71.14	9.06	82.13	6.73
S plantar C 150	54.28	5.87	57.47	4.40
S plantar E 50	177.16	15.15	202.69	11.17
S plantar E 150	156.12	13.55	161.16	10.08
We plantar C 50	56.42	8.35	69.00	6.20
We plantar C 150	40.72	4.79	46.24	3.59
We plantar E 50	143.10	15.92	170.23	11.74
We plantar E 150	119.64	13.85	132.16	10.31
S quadriceps C 50	88.54	8.43	103.09	7.23
S quadriceps C 150	76.01	6.09	80.84	4.97
S quadriceps E 50	148.57	10.87	159.44	8.98
S quadriceps E 150	143.64	9.42	157.06	7.69
We quadriceps C 50	69.19	8.14	84.95	6.98
We quadriceps C 150	64.52	5.94	64.79	4.84
We quadriceps E 50	123.47	9.56	136.52	7.90
We quadriceps E 150	124.47	9.04	135.40	7.37
S hamstrings C 50	35.71	9.92	43.93	3.46
S hamstrings C 150	41.05	3.16	47.23	2.65
S hamstrings E 50	58.83	4.61	68.19	3.87
S hamstrings E 150	65.50	3.51	69.99	2.87
We hamstrings C 50	26.25	3.77	33.31	3.24
We hamstrings C 150	30.42	3.25	37.28	2.72
We hamstrings E 50	43.67	4.19	56.61	3.51
We hamstrings E 150	55.12	3.21	60.07	2.63

Table A2.6 Data of faller and non-faller subjects for isokinetic strength normalised to square of corrected height. Dorsi=dorsiflexion, Plantar=plantarflexion S=strong, We=weak, C=concentric, E=eccentric. Numbers refer to speed.

Appendix 3. Data for comparison of frequent fallers (>3 falls/year) v non-fallers

	Frequent fallers		Non-fallers	
	Mean	Std. Error	Mean	Std. Error
Dorsi 30s	215	13	232	6
Dorsi 30we	154	15	196	7
Dorsi 20s	184	14	202	7
Dorsi 20we	138	13	170	6
Dorsi 10s	148	13	148	6
Dorsi 10we	115	15	108	7
Dorsi 0s	93	13	79	8
Dorsi 0we	59	9	47	5
Plantar 30s	156	14	218	15
Plantar 30we	126	17	169	14
Plantar 20s	248	21	354	20
Plantar 20we	211	21	287	19
Plantar 10s	381	52	512	25
Plantar 10we	337	51	422	25
Plantar 0s	469	67	624	34
Plantar 0we	422	67	520	34
Quadriceps s 90s	341	42	370	21
Quadriceps s 90we	299	39	324	19
Quadriceps s 80s	351	42	400	22
Quadriceps s 80we	305	38	353	19
Quadriceps s 70s	346	36	400	19
Quadriceps s 70we	309	36	360	18
Quadriceps s 60s	325	31	376	16
Quadriceps s 60we	285	31	340	16
Quadriceps s 50s	290	26	341	14
Quadriceps s 50we	251	26	303	13
Quadriceps s 40s	245	22	294	11
Quadriceps s 40we	212	21	257	11
Quadriceps s 30s	203	18	244	9
Quadriceps s 30we	178	17	210	8
Quadriceps s 20s	156	16	174	8
Quadriceps s 20we	142	16	168	8
Hamstrings 90s	109	15	123	7
Hamstrings 90we	89	14	103	6
Hamstrings 80s	109	13	135	6
Hamstrings 80w	89	13	113	6
Hamstrings 70s	113	13	144	7
Hamstrings 70w	91	13	121	6
Hamstrings 60s	124	15	152	8
Hamstrings 60w	102	14	129	7
Hamstrings 50s	125	16	157	8
Hamstrings 50w	101	15	131	8
Hamstrings 40s	127	17	161	9
Hamstrings 40w	110	17	136	9
Hamstrings 30s	144	19	172	9
Hamstrings 30w	112	20	144	10
Hamstrings 20s	140	21	172	10
Hamstrings 20w	113	21	144	10

Table A3.1 Isometric strength in frequent fallers and non-fallers (N). Dorsi=Dorsiflexion, Plantar=plantarflexion, S=strong, We=weak, numbers refer to angle.

	Frequent fallers		Non-fallers	
	Mean	Std. Error	Mean	Std. Error
S dorsi C 50	89	11	90	5
We dorsi C 50	70	8	73	4
S dorsi C 150	68	3	64	3
We dorsi C 150	61	3	55	2
S dorsi E 50	252	23	274	11
We dorsi E 50	230	23	237	11
S dorsi E 150	206	24	217	10
We dorsi E 150	184	24	185	10
S plantar C 50	147	17	231	18
We plantar C 50	110	15	192	16
S plantar C 150	113	12	158	10
We plantar C 150	87	9	127	9
S plantar E 50	431	37	578	28
We plantar E 50	341	48	486	27
S plantar E 150	360	44	455	22
We plantar E 150	287	39	377	24
S quadriceps C 50	290	43	316	23
We quadriceps C 50	226	43	263	23
S quadriceps C 150	242	30	246	16
We quadriceps C 150	207	30	199	16
S quadriceps E 50	458	51	487	27
We quadriceps E 50	387	46	414	25
S quadriceps E 150	454	44	478	23
We quadriceps E 150	388	45	409	23
S hamstrings C 50	120	20	137	11
We hamstrings C 50	97	20	105	11
S hamstrings C 150	144	16	144	9
We hamstrings C 150	106	17	115	9
S hamstrings E 50	190	23	209	12
We hamstrings E 50	148	21	173	11
S hamstrings E 150	213	18	212	9
We hamstrings E 150	180	16	181	8

Table A3.2 Isokinetic strength in frequent fallers and non-fallers (N). Dorsi=Dorsiflexion, Plantar=plantarflexion, S=strong, We=weak, C=concentric, E=eccentric, numbers refer to speed.

	Frequent fallers		Non-fallers	
	Mean	Std. Error	Mean	Std. Error
Quadriceps 90	14.12	3.62	12.46	1.16
Quadriceps 80	13.31	2.56	11.36	1.26
Quadriceps 70	13.90	3.17	10.67	1.18
Quadriceps 60	15.17	3.32	9.34	1.22
Quadriceps 50	13.94	3.81	10.28	1.60
Quadriceps 40	17.44	5.86	12.07	1.27
Quadriceps 30	12.10	3.40	12.75	1.63
Quadriceps 20	18.50	5.64	11.76	1.67
Hamstrings 20	19.27	3.06	16.42	2.04
Hamstrings 30	21.49	4.82	14.81	2.06
Hamstrings 40	13.05	4.99	16.53	2.06
Hamstrings 50	21.47	4.56	17.23	2.34
Hamstrings 60	29.42	8.60	15.11	1.73
Hamstrings 70	24.55	5.69	16.85	1.85
Hamstrings 80	24.15	7.50	17.46	2.11
Hamstrings 90	21.22	4.89	17.51	2.57
Dorsi 30	28.98	6.00	15.28	1.97
Dorsi 20	29.57	6.30	17.31	1.96
Dorsi 10	22.56	8.00	28.72	3.02
Dorsi 0	35.21	9.11	34.67	6.61
Plantar 0	13.10	3.39	19.88	2.69
Plantar 10	17.80	12.01	19.30	15.50
Plantar 20	15.56	2.90	18.15	2.14
Plantar 30	21.40	18.4	22.6	16.2
Quadriceps C 50	22.19	5.86	17.29	2.25
Quadriceps E 50	12.52	3.15	14.04	2.39
Quadriceps C 150	14.74	2.76	20.57	2.63
Quadriceps E 150	12.34	2.30	13.83	2.44
Hamstrings C 50	26.07	7.76	27.52	3.76
Hamstrings E 50	19.60	3.78	15.06	1.90
Hamstrings C 150	31.62	6.56	21.74	2.86
Hamstrings E 150	12.89	2.37	13.13	1.53
Dorsi C 50	16.84	3.18	18.99	2.69
Dorsi E 50	8.54	1.76	15.37	2.26
Dorsi C 150	10.29	3.27	13.90	2.13
Dorsi E 150	12.22	6.20	16.29	2.21
Plantar C 50	23.69	5.04	17.00	2.17
Plantar E 50	23.09	5.98	16.42	2.47
Plantar 150w	21.44	4.20	18.18	2.36
Plantar E 150	20.97	3.31	17.71	3.01
Power	23.48	5.01	13.88	1.47
Rf csa	14.20	5.78	16.64	1.90

Table 3.3 Asymmetry in frequent fallers and non-fallers. Dorsi=Dorsiflexion, Plantar=plantarflexion, C=concentric, E=eccentric, isometric numbers refer to angle, isokinetic numbers refer to speed.

	Frequent fallers		Non-fallers	
	Mean	Std. Error	Mean	Std. Error
POWER most powerful leg	178	20.24	180	9.84
POWER least powerful leg	153.9	20.79	157.3	10.20
RF CSA larger leg	4.2	0.73	4.6	0.29
RF CSA smaller leg	3.7	0.57	3.8	0.22

Table 3.4 Power and RF CSA in frequent fallers and non-fallers.

	Frequent fallers		Non-fallers	
	Mean	Std. Error	Mean	Std. Error
S quadriceps iso	0.08	0.02	0.12	0.03
We quadriceps iso	0.27	0.15	0.17	0.05
S hamstrings iso	0.20	0.06	0.14	0.02
We hamstrings iso	0.68	0.26	0.24	0.05
S dorsi iso	0.26	0.08	0.17	0.02
We dorsi iso	0.27	0.08	0.23	0.03
S plantar iso	0.14	0.03	0.16	0.03
We plantar iso	0.31	0.05	0.19	0.03
S quadriceps C	0.35	0.15	0.21	0.07
S quadriceps E	0.20	0.10	0.15	0.04
We quadriceps C	0.11	0.02	0.12	0.02
We quadriceps e w	0.11	0.03	0.12	0.03
S hamstrings C	0.17	0.04	0.22	0.04
S hamstrings E	0.27	0.06	0.17	0.04
We hamstrings C	0.27	0.18	0.21	0.04
We hamstrings E	0.27	0.09	0.28	0.06
S dorsi C	0.32	0.11	0.25	0.03
S dorsi E	0.33	0.10	0.21	0.02
We dorsi E	0.52	0.14	0.25	0.03
We dorsi E	0.52	0.31	0.26	0.03
S plantar C	0.20	0.06	0.18	0.04
S plantar E	0.20	0.06	0.20	0.05
We plantar C	0.20	0.04	0.20	0.04
We plantar E	0.34	0.03	0.17	0.02

Table 3.5 Antagonist co-activation in frequent fallers and non-fallers. Dorsi=Dorsiflexion, Plantar=plantarflexion, S=strong, We=weak, C=concentric, E=eccentric, Iso=isometric

Note: Values for non-fallers in this analysis may differ slightly from those in the original (fallers defined as having ≥ 1 fall) analysis as the exclusion of those fallers with 1-2 falls changed the sex make-up of the whole sample, thus altering the sex-correction factor.

Appendix 4 – Mathcad template for calculation of acceleration power at 6 frequency bands during functional tasks

$a_0 :=$

	0
0	
1	
2	

$a_1 :=$

	0
0	
1	

$$lim := (0\ \ 9\ \ 14\ \ 24\ \ 50\ \ 70\ \ 100)^T$$

$$n := 0..\,\mathrm{last}(a) \qquad \frac{\left(lim^T + 1\right)\cdot 200}{250} = \blacksquare \qquad q := 0..\,\mathrm{rows}(lim) - 2$$

$$fa_n := \mathrm{cfft}\left[\left(a_n\right)^{\langle 1\rangle}\right]$$

$$Bin_{n,q} := \sum_{k = lim_q}^{lim_{q+1}} \left|\left(fa_n\right)_k\right| \qquad Bin$$

$$r := 0..\,1000$$

$$\mathrm{rows}\left(a_n\right) =$$

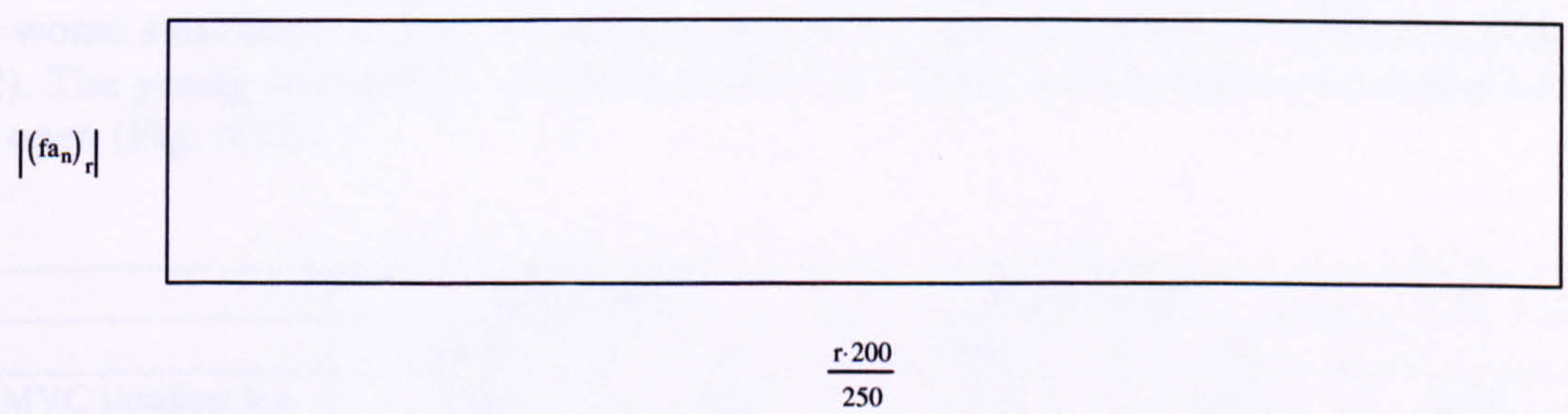
$$\blacksquare \qquad scale := \frac{200}{\mathrm{rows}\left(a_0\right)}$$

value of j corresponding to a frequency f is given by $\quad j(f) := \mathrm{round}\left(\frac{f}{scale}\right)$

$$Bins := (1\ \ 4\ \ 8\ \ 12\ \ 18\ \ 32\ \ 45)^T$$

$$jbin := \overrightarrow{j(Bins)} \qquad jbin^T = \blacksquare$$

$$Bin_{n,q} := \sum_{k = jbin_q}^{jbin_{q+1}} \left|\left(fa_n\right)_k\right| \qquad Bin = \blacksquare$$



Appendix 5. The effects of visual feedback on isometric force CoV in young and older subjects.

Introduction

Visual feedback may influence isometric steadiness (Kakuda et al. 1999, De Serres et al 2000). Since visual feedback of force levels are not used during functional activities, measurement of isometric steadiness without such feedback may be more valid.

Method

15 older and 12 young subjects derived from the young and older non faller groups (see chapter 2) were recruited. Isometric steadiness measurements were carried out on both legs at 10, 25 and 50% MVC as described in chapter 5. However, in addition to the trials with eyes open, a further trial with the monitor covered was carried out. Subjects were allowed to use the force output graph on the monitor screen to attain the correct force level, and after 2 seconds the monitor screen was covered until the end of the 10 second contraction. Subjects were asked to maintain the force level as steadily as possible. Only the middle 6 second portion was analysed. The order of eyes open/eyes closed was randomized for each force level, and both legs were measured in succession at the same condition. Paired t tests were used to compare measurements made with visual feedback to measurements made without at each contraction condition on the steadier and less steady legs. Independedent t tests were used to compare the age group difference in the change between CoV measured without and with visual feedback

Results

The young subjects showed worse steadiness without visual feedback at 25 and 10% MVC on the less steady leg (Fig. A5.1). The older subjects showed a similar pattern, with worse steadiness at 25% on the less steady leg and 10% MVC on both legs (Fig. A5.2). The young and old did not differ in the CoV difference between eyes closed and eyes open (Fig. A5.3)

	Eyes open		Eyes closed		P
	mean	sd	mean	sd	
50%MVC steadiest leg	1.58	0.42	1.7	0.67	0.58
50%MVC less steady leg	1.93	0.41	2.23	0.84	0.29
25%MVC steadiest leg	1.46	0.49	1.92	0.91	0.13
25%MVC less steady leg	1.7	0.59	2.83	1.54	0.03
10%MVC steadiest leg	1.45	0.63	1.72	0.62	0.31
10%MVC less steady leg	1.79	0.68	2.72	0.9	0.01

Fig. A5.1 CoV of isometric force in young subjects with and without visual feedback at 3 different force levels on the steadier and less steady legs.

	Eyes open		Eyes closed		P
	mean	sd	mean	sd	
50%MVC steadiest leg	1.48	0.51	1.83	0.93	0.22
50%MVC less steady leg	2.02	0.58	2.43	1.24	0.26
25%MVC steadiest leg	1.29	0.39	1.48	0.65	0.34
25%MVC less steady leg	1.76	0.8	2.56	1.29	0.05
10%MVC steadiest leg	1.16	0.39	2.02	1.12	0.01
10%MVC less steady leg	1.87	0.77	3.51	2.56	0.03

Fig. A5.2 CoV of isometric force in older subjects with and without visual feedback at 3 different force levels on the steadier and less steady legs.

	closed - open diff young		closed - open diff older		P
	mean	sd	mean	sd	
50%MVC steadiest leg	0.13	0.46	0.22	1.1	0.77
50%MVC less steady leg	0.29	0.56	0.25	1.7	0.93
25%MVC steadiest leg	0.46	0.69	0.19	0.53	0.25
25%MVC less steady leg	1.1	0.8	1.16	1.48	0.54
10%MVC steadiest leg	0.27	0.52	0.86	1.1	0.10
10%MVC less steady leg	0.93	0.86	1.66	2.48	0.25

Fig. A5.3 Difference between young and older subjects in change in CoV between no visual feedback and visual feedback at 3 different force levels on the steadier and less steady legs.

Discussion

Though these results do suggest that visual feedback affects steadiness at lower intensities in both young and older subjects, it should be noted that this was largely because subjects were unable to maintain the required force levels without visual reference to the monitor. This was particularly noticeable at lower intensities, where presumably proprioceptive input was lower. This tailing-off of traces artificially increased the measured CoV even though the actual force fluctuations were not noticeably affected on visual inspection. In addition, the age groups did not differ in this effect, suggesting visual feedback would not confound any age group comparison of steadiness. Thus to avoid the problems associated with force traces tailing off, to permit comparison with other studies using visual feedback, and because a large number of subjects had already been tested using visual feedback it was decided to continue to use visual feedback in the study.

Appendix 6: Test-retest reliability of steadiness tests

Introduction

Three types of steadiness test have been used in this study: isometric, anisometric and functional. There are no evaluations of these reliability tests in the literature as far as is known. Test retest reliability testing was therefore carried out as follows.

Methods

8 young subjects between the ages of 22 and 37 (mean 26.4, SE 3.2) were recruited from the young group in this study (Chapter 2). They were familiarised with the isometric, anisometric and functional steadiness test procedures using the equipment and procedure described in Chapter 5, and practice efforts were performed. Subjects then performed each of the three types of steadiness measure using their less dominant leg.

The measures were:

Isometric steadiness:	Force CoV at 50%MVC, Force CoV at 25%MVC, Force CoV at 10%MVC
Anisometric steadiness:	SD of acceleration during concentric 1kg contraction SD of acceleration during concentric 5kg contraction SD of acceleration during eccentric 1kg contraction SD of acceleration during eccentric 5kg contraction
Functional:	SD of knee angular acceleration during standing SD of knee angular acceleration during sitting SD of knee angular acceleration during stepping up SD of knee angular acceleration during stepping down

For isometric steadiness 2 MVC contractions were initially produced, and the better value was used to calculate the % MVC target levels. During the session each measure was tested twice, and the best value was recorded. This session was repeated three times, with between 2 and 7 days between each test, at the same time of day.

For each subject, the co-efficient of variation (CoV) was derived from the three repeated tests and Least Significant Difference (LSD) from the first two repeated tests.

CoV was calculated as the SD of values across the three repeated tests divided by the mean across the three repaeated tests. The mean CoV of all subjects was calculated, which represented the reliability measure.

For each measure, the first two sets of tests were paired for each subject, and the Least Significant Difference (LSD) for each measure was calculated. LSD = the standard deviation of the differences between the two tests x t, where t is the 2 tailed t value at the 5% significance level and n-1 degrees of freedom (n=8).

Results

LSD values were as shown in Table A6.1:

Measure	n	LSD
Stand	8	0.83
Sit	8	2.90
Step up	8	1.75
Step down	8	1.85
C 1	7	.0204
E1	7	.0200
C 5	7	.0064
E 5	7	.0076
CV 50%MVC	8	.0067
CV 25% MVC	8	.0065
CV 10% MVC	8	.0080

Table A6.1 LSD values for the steadiness measures

CoV values were as shown in Table A6.2 :

Measure	n	CoV (%)
Stand	8	7.2
Sit	8	8.4
Step up	8	8.7
Step down	8	8.4
C 1	7	11.6
Eccl	7	15.7
C 5	7	16.3
E 5	7	15.7
Force CoV 50%MVC	8	15.4
Force CoV 25% MVC	8	23.05
Force CoV 10% MVC	8	18.8

Table A6.2 Mean CoV values for the steadiness measures.

Discussion

The chief advantage of LSD is that it provides a measure of test retest reliability in the units of the measurement itself. The LSD value is therefore the minimum difference between readings that can be regarded as due to the independent variable.

Use of the CoV as a measure of reliability has been criticised because it is dependent on the absolute values of the measurement, which may confound results if different subjects vary in the measure (Stokes et al. 1997). However, since this confounding will always tend to over-estimate variability it acts as a conservative measure (Stokes et al. 1997).

Only young subjects were tested because of difficulties recruiting enough older subjects willing to attend for repeated tests. However there are indications that the tests would have been of similar reliability with older subjects. Although measures of inter-subject variability, such as SE or SD are partially an indicator of the normally distributed biological variability of the variable they are also influenced by the variability (reliability) of the measurement itself. If it is assumed that biological variability of young and older subjects was similar, the similar SE values between young and old for most variables (when scaled to the mean values) suggest that reliability was broadly similar.

Appendix 7 – Effects of strength training on steadiness.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
50% MVC steadiest leg	31	1.3 (0.1)	1.7 (0.1)	20	1.3 (0.9)	1.9 (0.2)
50% MVC less steady leg	32	2.0 (0.2)	1.7 (0.1)	19	2.0 (0.2)	1.6 (0.1)
25% MVC steadiest leg	33	1.2 (0.1)	1.7 (0.2)	22	1.1 (0.2)	1.8 (0.2)
25% MVC less steady leg	32	2.0 (0.2)	1.7 (0.2)	22	2.5 (0.3)	1.9 (0.2)
10% MVC steadiest leg	30	1.2 (0.2)	1.9 (0.3)	20	1.2 (0.1)	2.2 (0.2)
10% MVC less steady leg	30	2.2 (0.3)	1.7 (0.2)	22	2.3 (0.3)	1.9 (0.2)
Average of all	30	1.6 (0.1)	1.7 (0.1)	19	1.8 (0.2)	1.8 (0.1)

Table A7.1 Pre and post-test measurements of CoV of isometric force in the exercise and control groups. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
Concentric 1kg	13	0.025 (0.005)	0.032 (0.002)	9	0.028 (0.003)	0.031 (0.004)
Eccentric 1kg	13	0.026 (0.005)	0.042 (0.005)	9	0.027 (0.006)	0.039 (0.004)
Concentric 5kg	12	0.027 (0.003)	0.027 (0.002)	10	0.031 (0.005)	0.031 (0.004)
Eccentric 5kg	12	0.028 (0.007)	0.027 (0.004)	10	0.024 (0.003)	0.040 (0.012)
Average of all	12	0.026 (0.004)	0.032 (0.003)	9	0.027 (0.004)	0.035 (0.004)

Table A7.2 Pre and post-test measurements of SD of acceleration in the exercise and control groups in the left leg. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
Stand steadiest leg	34	6.4 (0.6)	8.3 (0.6)	24	5.9 (0.7)	7.1 (0.7)
Stand less steady leg	32	11.3 (1.5)	9.6 (1.5)	22	8.7 (1.4)	9.1 (1.4)
Sit steadiest leg	34	5.9 (0.4)	7.9 (0.6)	24	5.1 (0.5)	7.0 (0.9)
Sit less steady leg	33	11.7 (2.2)	6.7 (0.5)	23	7.2 (0.9)	7.6 (1.1)
Step up steadiest leg	33	6.9 (0.4	8.1 (0.4)	24	6.2 (0.5)	7.7 (0.5)
Step up less steady leg	32	11.1 (1.4)	8.8 (0.6)	23	11.3 (1.0)	8.6 (1.0)
Step down steadiest leg	32	7.1 (0.6)	8.1 (0.5)	23	6.9 (0.7)	8.5 (0.8)
Step down less steady leg	33	11.5 (1.4)	8.6 (0.6)	24	11.0 (1.0)	8.7 (1.1)
Average of all	30	9.1 (0.7)	8.2 (0.5)	20	7.4 (0.8)	8.1 (0.6)

Table A7.3 Pre and post-test measurements of standard deviation of knee angular acceleration in the exercise and control groups. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean(SE)	Mean(SE)		Mean(SE)	Mean(SE)
Step up steadiest 1-4Hz	32	105.9 (6.3)	129.5 (4.4)	22	104.2 (8.1)	105.3 (5.3)
Step up steadiest 4-8Hz	32	69.4 (4.7)	91.7 (7.0)	22	73.5 (8.4)	80.0 (5.5)
Step up steadiest 8-12Hz	32	51.9 (3.7)	60.8 (7.4)	22	51.8 (5.3)	55.2 (5.3)
Step up steadiest 12-18Hz	32	55.6 (4.7)	60.3 (9.7)	22	48.1 (5.1)	59.8 (9.5)
Step up steadiest 18-32Hz	32	91.2 (9.2)	85.1 (9.9)	22	69.9 (11.1)	76.3 (11.9)
Step up steadiest 32-45Hz	32	26.8 (2.9)	25.5 (2.3)	22	20.8 (1.4)	24.3 (2.7)
Step up less steady 1-4Hz	30	144.3 (7.8)	136.0 (5.7)	21	139.9 (9.3)	118.6 (6.8)
Step up less steady 4-8Hz	See Table 9.4					
Step up less steady 8-12Hz	30	110.4 (17.4)	51.8 (4.4)	21	113.9 (29.3)	62.5 (9.9)
Step up less steady 12-18Hz	30	165.7 (40.2)	51.0 (4.7)	21	122.6 (21.9)	57.3 (12.5)
Step up less steady 18-32Hz	30	274.0 (54.9)	72.8 (5.8)	21	191.6 (27.9)	89.8 (18.8)
Step up less steady 32-45Hz	30	83.3 (22.2)	21.8 (1.4)	21	41.4 (4.8)	24.8 (3.4)

Table A7.4 Pre and post-test measurements of acceleration power (rad.sec $^{-2}$)^2 of knee angular acceleration in 6 spectral bands during the step up manouvre in the exercise and control groups. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
Step down steadiest 1-4Hz	31	100.2 (5.5)	104.1 (5.0)	21	98.7 (6.7)	103.2 (6.1)
Step down steadiest 4-8Hz	31	74.4 (5.1)	83.0 (6.7)	21	70.1 (7.5)	90.9 (7.4)
Step down steadiest 8-12Hz	31	54.9 (3.7)	53.5 (4.8)	21	53.7 (5.4)	60.3 (5.7)
Step down steadiest 12-18Hz	31	58.5 (7.0)	50.5 (4.2)	21	50.1 (6.8)	55.4 (8.0)
Step down steadiest 18-32Hz	31	99.3 (17.1)	77.9 (7.9)	21	75.9 (8.7)	79.2 (15.6)
Step down steadiest 32-45Hz	31	27.3 (4.2)	25.0 (2.3)	21	23.7 (1.8)	24.5 (4.3)
Step down less steady 1-4Hz	31	127.4 (9.9)	113.2 (5.4)	22	140.8 (11.6)	108.5 (6.4)
Step down less steady 4-8Hz	31	113.9 (10.5)	92.2 (5.6)	22	115.4 (9.3)	103.7 (12.1)
Step down less steady 8-12Hz	31	103.5 (13.1)	64.8 (6.4)	22	93.3 (11.1)	65.0 (10.0)
Step down less steady 12-18Hz	31	163.1 (34.3)	62.6 (6.7)	22	124.0 (18.6)	73.3 (21.1)
Step down less steady 18-32Hz	31	272.6 (50.9)	101.5 (12.7)	22	192.9 (26.6)	118.9 (31.5)
Step down less steady 32-45Hz	31	69.1 (9.7)	32.1 (3.9)	22	57.8 (7.6)	33.1 (5.4)

Table A7.5 Pre and post-test measurements of acceleration power (rad.sec $^{-2}$)^2 of knee angular acceleration in 6 spectral bands during the step down manouvre in the exercise and control groups. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
Stand steadiest 1-4Hz	33	83.4 (5.8)	112.6 (7.7)	19	84.1 (8.4)	98.2 (10.3)
Stand steadiest 4-8Hz	33	73.2 (4.7)	76.8 (5.8)	19	79.3 (9.3)	65.2 (5.9)
Stand steadiest 8-12Hz	33	68.4 (6.9)	69.6 (7.2)	19	57.3 (6.7)	63.8 (11.4)
Stand steadiest 12-18Hz	33	86.9 (8.5)	98.8 (13.4)	19	75.5 (7.7)	95.8 (21.6)
Stand steadiest 18-32Hz	33	160.1 (19.4)	171.8 (33.5)	19	124.3 (18.7)	174.5 (57.0)
Stand steadiest 32-45Hz	33	47.6 (4.2)	47.6 (7.4)	19	40.7 (6.2)	81.1 (42.9)
Stand less steady 1-4Hz	32	108.8 (9.1)	104.6 (5.2)	19	105.4 (10.3)	93.8 (8.7)
Stand less steady 4-8Hz	32	120.7 (12.6)	82.9 (8.7)	19	111.2 (15.3)	77.4 (10.2)
Stand less steady 8-12Hz	32	120.6 (13.6)	94.2 (17.4)	19	97.7 (9.9)	76.8 (12.7)
Stand less steady 12-18Hz	32	224.0 (51.2)	131.3 (26.6)	19	127.7 (13.2)	123.6 (26.0)
Stand less steady 18-32Hz	32	377.0 (86.5)	202.4 (41.9)	19	239.2 (43.1)	187.0 (35.1)
Stand less steady 32-45Hz	32	108.9 (13.6)	57.1 (6.8)	19	85.9 (19.0)	53.7 (8.9)

Table A7.6 Pre and post-test measurements of acceleration power $(rad.sec^{-2})^2$ of knee angular acceleration in 6 spectral bands during the stand manouvre in the exercise and control groups. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
Sit steadiest 1-4Hz	33	79.3 (4.6)	89.2 (5.7)	19	72.2 (8.3)	82.6 (8.5)
Sit steadiest 4-8Hz	33	79.1 (6.0)	105.9 (13.4)	19	74.2 (7.0)	78.6 (9.6)
Sit steadiest 8-12Hz	33	65.3 (5.1)	82.1 (10.1)	19	64.5 (9.2)	83.8 (16.7)
Sit steadiest 12-18Hz	33	79.5 (6.5)	105.8 (13.8)	19	61.3 (5.5)	110.2 (19.7)
Sit steadiest 18-32Hz	33	135.2 (10.2)	211.4 (35.6)	19	110.5 (10.3)	187.3 (38.4)
Sit steadiest 32-45Hz	33	42.6 (3.1)	61.8 () (8.4)	19	43.7 (4.8)	56.0 (9.9)
Sit less steady 1-4Hz	32	98.6 (6.6)	97.4 (8.6)	19	85.9 (6.2)	92.1 (8.1)
Sit less steady 4-8Hz	32	114.3 (12.2)	93.4 (7.4)	19	107.5 (11.2)	87.6 (10.3)
Sit less steady 8-12Hz	32	131.5 (24.4)	76.4 (6.9)	19	99.3 (13.8)	70.5 (7.3)
Sit less steady 12-18Hz	32	222.4 (51.5)	89.8 (8.4)	19	122.5 (19.3)	97.3 (15.9)
Sit less steady 18-32Hz	32	351.9 (71.4)	171.5 (20.9)	19	217.2 (31.0)	196.2 (42.6)
Sit less steady 32-45Hz	32	107.1 (17.4)	60.4 (9.1)	19	72.0 (8.0)	62.9 (14.1)

Table A7.7 Pre and post-test measurements of acceleration power $(rad.sec^{-2})^2$ of knee angular acceleration in 6 spectral bands during the sit manouvre in the exercise and control groups. There were no group x time interactions for any variables.

	EXERCISE			CONTROL		
	N	PRE	POST	N	PRE	POST
		Mean (SE)	Mean (SE)		Mean (SE)	Mean (SE)
Average 1-4Hz	30	106 (6)	109 (4)	19	102 (8)	99 (5)
Average 4-8Hz	30	98 (6)	92 (5)	19	93 (8)	83 (6)
Average 8-12Hz	30	91 (5)	66 (4)	19	80 (8)	68 (6)
Average 12-18Hz	30	135 (15)	83 (6)	19	94 (2)	86 (8)
Average 18-32Hz	30	230 (24)	138 (5)	19	159 (31)	140 (16)
Average 32-45Hz	30	70 (6)	42 (4)	19	50 (8)	46 (5)

Table A7.8 Pre and post-test measurements of acceleration power $(rad.sec^{-2})^2$ of knee angular acceleration in 6 spectral bands averaged across tasks and legs in the exercise and control groups. There were no group x time interactions for any variables.

	Exercise			Control		
Asymmetry	n	pre	post	n	pre	post
		Mean(SE)	Mean(SE)		Mean(SE)	Mean(SE)
Standing steadiness	32	0.28 (0.04)	0.19 (0.04)	22	0.26 (0.04)	0.25 (0.04)
Sitting steadiness	33	0.29 (0.04)	0.21 (0.03)	23	0.24 (0.04)	0.25 (0.04)
Stepping up steadiness	32	0.27 (0.04)	0.19 (0.03)	23	0.42 (0.04)	0.24 (0.04)
Stepping down steadiness	32	0.30 (0.04)	0.19 (0.02)	23	0.30 (0.05)	0.30 (0.05)
50% MVC steadiness	28	0.26 (0.03)	0.24 (0.03)	18	0.32 (0.05)	0.26 (0.05)
25% MVC steadiness	32	0.31 (0.04)	0.29 (0.04)	22	0.36 (0.05)	0.38 (0.05)
10% MVC steadiness	30	0.39 (0.04)	0.40 (0.05)	20	0.41 (0.05)	0.40 (0.06)
Average of above	24	0.30 (0.02)	0.25 (0.02)	16	0.34 (0.03)	0.29 (0.02)

Table A7.9 Asymmetry in the isometric and functional steadiness. There were no group x time interactions.

	EXERCISE			CONTROL		
	PRE		POST	PRE		POST
	N	Mean(SE)	Mean(SE)	N	Mean(SE)	Mean(SE)
Quads 90°	26	12.7 (1.7)	13.9 (1.8)	16	13.4 (1.7)	15.0 (2.9)
Quads 80°	27	13.7 (2.1)	12.6 (2.1)	14	10.5 (1.6)	10.9 (2.1)
Quads 60°	27	10.0 (1.4)	9.5 (1.5)	17	8.4 (2.1)	11.0 (1.7)
Quads 50°	26	12.5 (2.1)	12.8 (1.0)	15	10.1 (2.7)	8.9 (1.8)
Quads 30°	26	15.4 (2.5)	11.4 (1.8)	15	11.6 (2.2)	9.2 (2.1)
Hams 30°	29	16.1 (2.7)	16.8 (2.3)	16	15.4 (3.4)	19.0 (4.1)
Hams 50°	See Table 9.4					
Hams 60°	26	20.4 (2.6)	12.4 (2.3)	16	17.8 (3.1)	13.1 (3.2)
Hams 80°	See Table 9.4					
Hams 90°	24	18.6 (3.6)	16.3 (2.3)	14	22.3 (3.8)	17.1 (2.8)
Average isometric	22	15.5 (1.4)	12.5 (1.1)	14	14.7 (1.8)	12.5 (1.4)

Table A7.10 Pre and post-test measurements of isometric strength symmetry in the exercise and control groups. There were no significant group x time interactions.

	EXERCISE			CONTROL		
	PRE		POST	PRE		POST
	N	Mean (SE)	Mean (SE)	N	Mean (SE)	Mean (SE)
Quads C 50°.sec^{-1}	22	16.7 (2.5)	16.3 (1.8)	12	24.4 (3.4)	13.2 (3.3)
Quads C 150 °.sec^{-1}	24	19.5 (2.7)	17.0 (2.5)	12	21.8 (3.6)	13.3 (3.3)
Hams C 50 °.sec^{-1}	23	26.2 (3.9)	22.0 (3.2)	12	23.7 (6.3)	14.8 (4.3)
Hams C 150 °.sec^{-1}	23	24.7 (3.9)	25.3 (3.9)	11	27.6 (5.2)	32.9 (6.3)
Average con	20	23.8 (2.1)	19.8 (2.3)	10	27.3 (2.8)	16.7 (3.1)
Quads E 50 °.sec^{-1}	21	14.9 (2.6)	14.0 (2.3)	12	10.6 (2.1)	9.9 (2.1)
Quads E 150 °.sec^{-1}	21	10.6 (1.6)	15.0 (2.3)	12	9.1 (1.9)	15.7 (2.8)
Hams E 50 °.sec^{-1}	24	17.1 (2.3)	16.6 (2.1)	12	20.8 (4.0)	13.9 (2.8)
Hams E 150 °.sec^{-1}	24	12.3 (1.7)	14.8 (2.1)	11	17.6 (2.5)	19.6 (4.4)
Average ecc	20	13.5 (1.3)	15.3 (1.3)	10	13.3 (1.9)	16.2 (1.8)
RF CSA (best/worst)	19	17.1 (3.1)	8.9 (1.5)	8	14.3 (3.0)	10.1 (2.9)

Table A7.11 Pre and post-test measurements of isokinetic strength and RF CSA symmetry in the exercise and control groups. There were no significant group x time interactions C=concentric, E=eccentric.

Appendix 8 - Biomechanical considerations

The quadriceps torque exerted at knee = quadriceps tensile force (f) x moment arm (d) (distance of axis of knee rotation to patella.

This will equal the torque measured at the ankle = force measured by force transducer at ankle (F) x shank length (D).

Therefore fd=FD

Therefore f/F=D/d

As D/d can be assumed to be fairly constant over a large group of individuals (because body dimensions tend to remain in a fairly constant proportion),

f/F=constant

Therefore f is proportional to F

Therefore the force at the ankle is a fair index of the quadriceps tensile force.